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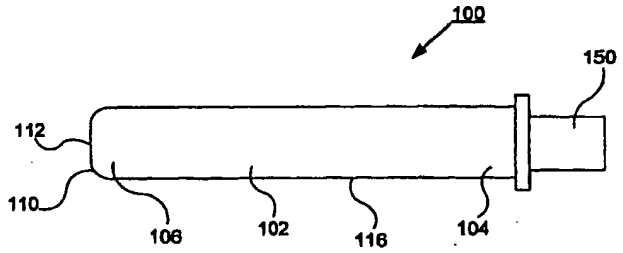
(54) Title: MEDICAL DEVICE UTILIZING HYDROGEL MATERIALS

(57) Abstract

The subject invention pertains to novel medical devices having highly advantageous physical characteristics which can reduce potential injury to patients. The excellent characteristics of the medical devices of the subject invention result, in part, from the use of hydrogel materials. Tubes for endotracheal devices can be formed entirely or partially of a swellable hydrogel material. The hydrogel material swells in the presence of water to form a seal within a tracheal lumen. In one embodiment, the endotracheal device is made fully of hydrogel. In another embodiment, the endotracheal device includes a hollow, pliable tube and a hydrogel sleeve attached to the tube. In another embodiment, the endotracheal device comprises a hollow, pliable tube, an inflatable cuff, and a hydrogel sleeve located over the cuff. In yet another embodiment, the endotracheal device comprises a hollow, pliable tube, and a hydrogel sleeve which acts as a true "gel–cuff". A further aspect of the subject invention concerns novel manufacturing processes used to produce novel hydrogel materials and the corresponding hydrogel materials. In addition, the subject invention relates to a method for delivery of treatments and chemicals substances to a patient.



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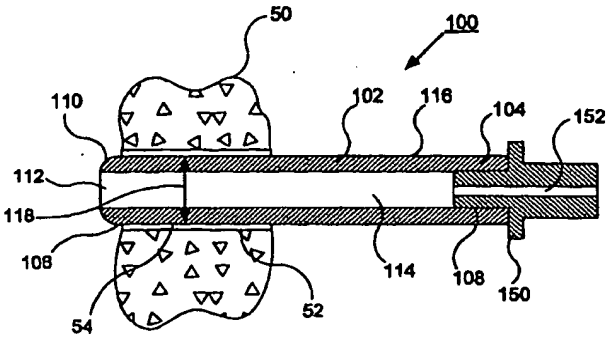

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DESCRIPTION

MEDICAL DEVICE UTILIZING HYDROGEL MATERIALS

Background of the Invention

The subject invention pertains to medical devices and related procedures utilizing hydrogel materials. In a specific embodiment, the present invention relates to endotracheal devices for use in, for example, short term emergency situations and long term chronic applications. An endotracheal device is a medical apparatus for use in connection with the pulmonary ventilation of a patient. Endotracheal devices are also used to administer anesthetics and to prevent the reflux of vomitus into a patient's lungs. The subject invention further pertains to methods for producing hydrogel materials and for bonding hydrogels to other materials.

More than twenty million procedures involving endotracheal intubation are performed each year in the United States. These procedures involve the placement of a tube into the tracheal lumen, and can be prescribed in any setting in which a patient airway must be established and controlled. The purpose of the endotracheal tube in these procedures is to maintain a viable airway, facilitate spontaneous and/or mechanical ventilation, allow the administration of inhalational agents, and/or reduce the reflux of vomitus into the lungs. In order to satisfy these requirements it is preferred to maintain a nearly airtight (5%-10% leak) seal between the tube and the tracheal wall. Most conventional endotracheal tubes provide this seal by employing a balloon or 'cuff' that is inflated once the tube is in place.

Typical endotracheal devices have a pliable, hollow tube with openings at opposite ends. The tube is inserted into a patient's trachea such that one end of the tube is located within the tracheal lumen midway between the vocal cords and the carina. The upper end of the tube is connected to a gas supply or breathing apparatus. In particular, the upper end may be provided with a friction fitting (normally a 15/22 mm standard connector) for attachment to a mechanical ventilating device.

Conventional endotracheal tubes are commonly constructed from transparent polyvinyl chloride (PVC) plastic and often feature a high-volume, low-pressure cuff, however, other designs are available as well. Prior art endotracheal devices with inflatable cuffs are illustrated in U.S. Patents Nos. 5,520,175 (Fry) and 5,329,921 (Socaris). The tube should be flexible to help minimize airway damage, but still rigid enough to ease intubation and provide airway security. Once inflated, the cuff can maintain the near seal and help to prevent fluid aspiration. The primary advantage of the high-volume, low pressure cuff design over previous devices is the increased contact area of the cuff. This feature is designed to decrease the forces applied to

the tracheal tissues by distributing the sealing pressure over a larger surface area. This approach should reduce the incidence of injury associated with cuff pressure. High volume, low-pressure endotracheal tubes represent the current state of the art for airway devices.

The condition of the lungs affects airway pressure and is considered by the physician in determining the method of anesthesia and ventilation. One method of characterizing lung condition is with compliance. Lung compliance describes the impedance of the lung to inflation, a highly compliant lung being one that is easily inflated at lower pressures. Most young and healthy people have lungs in this condition. However, lung compliance decreases with age and infirmity, so that an elderly or pulmonary injured patient will normally have lower compliance or 'stiff' lungs. Lungs may also become stiff with fluid accumulation, pneumonia, or trauma. In these situations the ventilation pressure must be increased to overcome the lung impedance, increasing the possibility of pressure related airway injuries.

Decreased lung compliance is not the only factor contributing to high airway pressure. The airway, starting at the glottic opening down the trachea to the terminal bronchioles can also contribute to the need for increased inflating pressure due to narrowing of the airways. Thus, lung compliance (C_L) and airway resistance (R_{AW}) can both contribute to impedance in the airway. In some instances, chest wall compliance (C_T) may also contribute.

To maintain airway pressure and ventilate the patient a seal must be created between the endotracheal tube and the tracheal lining. To maintain this seal with minimal leakage the pressure in the cuff must equal or exceed the airway pressure. This is problematic because the cuff pressure is exerted directly on the tracheal lining. Since it has been clinically established that high cuff-to-tracheal wall (CT) pressures contribute to tracheal wall ischemia, necrosis, wall erosion, tracheal-esophageal fistula, or other tissue damage, it follows that injury to these tissues is more likely to occur with increased impedance to ventilation. This is because greater airway pressures require greater CT pressures to maintain adequate seal. Furthermore, because these patients are often stressed, elderly, or immuno-comprised, it is very easy for life-threatening illnesses to develop. The presence of cuff-related tracheal injury is particularly troublesome because it creates a pathway for secondary infection. Ventilator-associated pneumonia is an unfortunate complication and frequent cause of death for many of these patients.

Conventional cuffs are abrasive and under certain conditions require high CT pressures to form and maintain satisfactory seals. Consequently, these cuffs are a source of irritation and sometimes cause serious injury to tracheal tissues. Injuries caused by inflatable cuffs range from severe sore throats to tracheal wall erosion.

The primary function of the trachea is to provide a passage between the larynx and the branching of the main stream bronchi. However, the trachea is not a simple conduit to the lungs,

but a complex structure composed of incomplete hyaline cartilage rings connected longitudinally by smooth muscle tissue. These cartilage rings are often C-shaped, with the end gap joined by the Trachealis muscle. This muscle regulates the gap and allows for the expansion and contraction of the trachea that occurs naturally during respiration. The overall framework is solid enough to provide mechanical integrity, maintain the roughly cylindrical shape of the lumen, and still allow the flexibility required by normal motion of the neck and head.

The most fragile and vulnerable component of the trachea is the mucosal lining. This overlays the basement membrane and is composed primarily of ciliated pseudo columnar epithelium and goblet cells. The goblet cells secrete a thick and watery mucous which keeps the surface insulated and moist, and also serves to entrap inhaled and resident particulate matter. The cephalid beating action of the cilia transports this foreign matter to the base of the larynx, where it is swallowed or expelled. The mucociliary transport system is part of the tracheal defense mechanism and prevents infectious agents from establishing a foothold in the trachea. Unfortunately, the mucociliary transport system can be severely compromised by conventional ET tube cuff designs and materials.

Complications with tracheal intubation range from mild epithelial desquamation to complete erosion of the cuff through the tracheal wall. These injuries are caused by a combination of the materials employed and the forces exerted by the cuff on the tracheal tissues. The abrasive action of the cuff shears cells from the lining, epithelium adhering to the cuff is removed during extubation, and normal forces exerted on the basement tissues disrupt the blood supply and cause pressure necrosis. These injuries tend to become more severe with increased intra-cuff pressure and the duration and magnitude of positive-pressure ventilation. The first areas affected are generally those of the mucosal membrane, since they are in intimate contact with the endotracheal tube. As these injuries penetrate deeper they become more serious and begin to affect the blood supply, the cartilage rings, and even the surrounding tissues.

The most elementary model of tracheal intubation assumes a cylindrical cuff placed statically in a cylindrical tracheal lumen. This is inaccurate however, because the trachea is neither cylindrical nor static. In reality, the human trachea may exhibit an unlimited variety of cross-sectional shapes including circular, ellipsoidal, or even triangular. The ability of air-inflated cuffs to conform to any of these shapes is limited. Furthermore, because conventional endotracheal tube cuffs have such well-defined circular profiles, the non-ideal tracheal lumen may be forced to conform to the shape of the cuff. This ultimately creates stress concentrations in the tissues that result in injury, especially at elevated airway pressures. The seriousness of this injury is often determined by the deviation of the tracheal cross-section from the ideal cuff cross-section.

Many clinicians are unaware of tracheal dynamics. This is of concern because the trachea continuously elongates, contracts, dilates, and expands during ventilation. In fact, current cuff designs cannot maintain seal under these conditions without exerting excessive pressure on the tracheal tissues. To accommodate expansion of the lumen the cuff must essentially be over-inflated. Once the cuff is in this state the trachea may dilate further, requiring additional cuff inflation to maintain seal. This series of events may continually repeat, leading to a 'ratcheting effect' in which the trachea is stretched to the elastic limits of the tissue.

In addition to the movement of the trachea, both rotational and axial relative motion occur between the cuff and tracheal lining during ventilation. Because common cuff materials are abrasive this results in a 'sanding effect' which may damage the lining. The potential for serious injury is significant, especially at higher airway pressures and during extended ventilation.

Damage to the mucosal membrane is related to both cuff design and material. Almost all conventional endotracheal tube cuffs are constructed from PVC, which is a relatively hydrophobic plastic. This type of material is resistant to water and generally tends to promote cellular adhesion once contact is made. A limited amount of pressure is often sufficient to breach the mucosal barrier and bring the cuff into direct contact with the lining. Once this occurs epithelial cells will often adhere to the cuff, especially if subjected to previous irritation. Relative motion may then forcibly remove these cells from the tissue bed during ventilation. Any adherent cells which survive ventilation will probably be injured or pulled out during extubation.

The surfaces of many commercially available endotracheal tubes have been shown to be contaminated with abrasive particulate matter. Ordinary vinyl cuffs are relatively abrasive to touch and microscopically rough as evidenced by scanning electron microscopy and atomic force microscopy. This type of surface can easily damage the tracheal tissues through the relative motion that occurs during ventilation. Even in the absence of adhesion phenomenon, the combination of particulate contamination and surface roughness is capable of causing significant abrasive injury. Cells previously irritated by adhesion are particularly vulnerable to further injury by this mechanism. These distressed cells may be sheared from the lining, leaving a gap in the tissue bed and damaging adjacent tissue.

Injury to the mucosa is irreversible because the new growth that occurs during recovery is in the form of cuboidal, not ciliated epithelium. In fact, if the injury is significant the post-recovery lining may show a complete absence of both goblet and ciliated pseudo columnar epithelial cells. Furthermore, even relatively mild cuff pressures may damage the mucosa. It has been estimated that CT pressures of approximately 10 mm- 20 mm Hg are sufficient to

impede blood flow to the tracheal tissues, and that any duration of intubation will lead to some level of permanent damage. When the cuff pressure is relatively low the tissue between the cartilage rings may partially escape compression, but in the region of the rings the compression is more severe and the blood flow is likely to be impaired. Electron photomicrographs invariably reveal the greatest tissue damage in this region. If blood flow is restricted for extended periods the surrounding tissues may be destroyed. Conditions requiring extended intubations at high inflation pressures are not uncommon. Blunt trauma to the airway or lungs may occur in, for example, a car accident. Fluid accumulation in the lungs is common for drowning victims.

Evidence that these complications are widely recognized is demonstrated in the number of solutions that have been developed. A variety of designs have been tried, including revised cuff designs, new materials, and exotic surface treatments. In addition, at least one radically different tube design is undergoing evaluation. Several variants have been suggested, including multiple cuffs, dual-walled cuffs, longer cylindrical cuffs, and foam-filled cuffs. The most recent innovation adopted by industry, the high-volume, low pressure cuff, was implemented more than twenty years ago. Each of these changes attempts to address one or more of the previously discussed complications, but for several reasons only a small portion of these ideas have been adopted by industry. Failure to demonstrate improvement in performance has prevented most of these devices from achieving widespread acceptance. Further, endotracheal tubes have commodity status, so high cost is also cited as a key factor in most cases.

A limited number of alternative materials are available on the market. Tubes manufactured from natural rubber, polyurethane, silicone rubber, and even polyethylene are available. Natural rubber tubes are being phased out, while polyurethane is available as a special order item. Silicone rubber tubes are available in commercial quantities, but are not in widespread use. The highly biocompatible and smooth surface presented by silicone should translate into fewer problems with abrasion and cellular adhesions. However, the high gas permeability and modest mechanical properties of silicone makes these tubes prone to leaks, and high-volume, low-pressure cuffs are not available. In addition, silicone tubes are also much more expensive than conventional tubes.

Several hydrophilic coatings applied by spray, dip-coating, or graft polymerization are mentioned in the patent literature. Some of these coatings are applied as a manufacturing step, while some are designed to be applied by the end-user to existing medical deices. Hydrophilic coatings are designed to improve general biocompatibiltiy by apposing a highly wettable surface to the target tissue. Coating processes often suffer from poor durability, inadequate thickness,

and also add considerable expense to a product. In addition, these products do not allow the endotracheal tube to adapt to tracheal geometry.

An entirely new tube design employing a series of 'ultra-thin gills' in place of the traditional cuff was recently proposed by scientists at the National Institutes of Health. This device is constructed from polyurethane elastomer and reinforced by nickel-alloy wire. The device is designed to be placed so that the gills 'straddle' the vocal cords in an attempt to maintain a seal at that location. Ideally, this device should prevent damage to the mucosal lining by creating and maintaining the seal external to the trachea in the vocal cords. Independent testing has not yet been disclosed, but possible drawbacks include poor seal at low-compliance, inadequate airway security, vocal cord irritation, high production costs, and increased training costs.

Brief Summary of the Invention

The subject invention pertains to novel medical devices having highly advantageous physical characteristics which reduce potential injury to patients. The excellent characteristics of the medical device of the subject invention result, in part, from the use of hydrogel materials. Hydrogels are a class of polymers which swell in the presence of moisture and may contain up to approximately 95% water by weight. In general these materials are very soft, very smooth, highly lubricious, non-abrasive, and non-adhesive to tissues. A further aspect of the subject invention concerns novel manufacturing processes to produce novel hydrogel materials and the corresponding hydrogel materials themselves. In addition, the subject invention relates to a method for delivery of chemical substances to a patient. A variety of techniques to produce hydrogel articles for use with the subject invention are known to those skilled in the art, including but not limited to, casting, molding, extrusion, pultrusion, and calendering.

In a specific embodiment, an endotracheal tube device (ETD) is formed essentially entirely of hydrogel material. In another specific embodiment, an ETD comprises a reinforcement which is covered with a sleeve formed essentially entirely of hydrogel material, such that the reinforcement enhances the structural integrity of the ETD. This reinforcement may be constructed from, for example, PVC, polyurethane (PU), polyethylene (PE), polypropylene (PP), nylon, wire mesh, or any other material with appropriate properties. Advantageously, volume swelling of the hydrogel material can allow a seal to be made between the ETD and a patient's trachea. Further embodiments of ETD's employ inflatable cuffs and/or gel-cuffs to allow quick sealing of the ETD with a patient's trachea. Additional embodiments of the subject invention relate to hydrogel sleeves which can be used in conjunction with devices

which contact human tissue during use, such that the use of hydrogel material reduces the potential for patient injury.

The subject invention pertains to a hydrogel sleeve or tube employed in the construction of the aforementioned devices . This tube or sleeve can be molded separately, in conjunction with, or in place over the underlying device. Advantageously, this tube or sleeve may be molded in a wide variety of cross-sectional shapes with or without a macroscopically smooth surface. In a specific embodiment an imprint or shape may be molded into or imparted to the surface. This shape might serve to, for example, enhance sealing, assist mucociliary transport, or ease device extraction.

The subject invention also relates to any tube, drain, catheter, device, or instrument, with or without a cuff or balloon, designed to convey fluids into or out from the body. The subject invention can be employed in any application involving the passage of gases or liquids into or out from the body, including but not limited to airway maintenance, drug delivery, dialysis, enteral feeding, and cavity drainage. The subject invention can be applied to any device which is inserted into the body and would benefit from the smooth, lubricious, non-abrasive, and non-adhesive surface provided by hydrogel materials. Specific applications include, but are not limited to, endotracheal tubes, tracheotomy tubes, nasopharyngeal tubes, nasogastric tubes, chest tubes, wound drains, intravenous catheters, burn treatment dressings, intravascular catheters, peritoneal dialysis catheters, and Foley catheters. Other applications include but are not limited to: angioplasty catheters, shunts, stents, endoscopes, laproscopes, and surgical laser devices. Furthermore, methods for delivery of drugs, chemical agents, or other substances can be accomplished utilizing the devices of the subject invention.

The subject invention pertains to a method which utilizes the devices of the subject invention for the delivery of chemical agents systemically or locally to specific tissues. For example, the integral hydrogel tube or sleeve can be loaded with or bonded to chemical agents including but not limited to antiseptics, antibiotics, anti-inflammatory agents, vectors, enzymes, and anti-thrombogenic agents. These agents can be chemically bonded to the device at any internal or external surface, or loaded into the bulk of the hydrogel. The hydrogel sleeves in these devices may also contain drug-loaded microspheres which release various agents over time, according to temperature, or in response to pH stimulus. Referring to Figures 2A and 2B, the Type-2 hybrid design has the additional ability to deliver agents from an internal cavity. Bulk loaded agent delivery rates may be controlled by manipulating the chemical composition of the delivered agent and hydrogel, as well as hydrogel morphology and porosity. Agent delivery rates for the Type-2 hybrid, see Figures 2A and 2B, are additionally affected by hydrogel water content, hydrogel porosity, drug concentration, and delivery pressure.

The subject invention also pertains to a class of interpenetrating polymer network (IPN) hydrogel materials found to be particularly useful in the construction of the full tube and hybrid tube designs of the subject invention. This class of IPN materials can be fabricated, for example, from PVP (polyvinyl pyrrolidone) and PHEMA (polyhydroxyethyl methacrylate) polymers in a variety of compositions to yield a wide array of properties, including remarkable lubricity, flexibility, and toughness.

The subject invention also relates to a process for producing PVA (polyvinyl alcohol) hydrogels found to be particularly useful in the construction of the aforementioned full tube and hybrid tube designs in accordance with the subject invention. This process is based on the well known freeze-thaw method for producing PVA hydrogels, but can involve the use of an additional dehydrating solvent to abbreviate the production cycle, for example to a single step. Dehydrating solvents, for example acetone, ethanol, or methanol, may be employed to remove water from the PVA gel after completion or prior to completion of the first freeze-thaw cycle to consolidate the remaining amorphous material and strengthen the gel. Advantageously, this process quickly produces hydrogels that are highly lubricious and very strong.

The subject invention also pertains to composite hydrogel materials fashioned from standard PVA hydrogels and freeze-thaw PVA hydrogels. These gels may be layered and bonded, for example, by means of adhesives, solvent bond, interference, heat, pressure, or any combination of the above. The combination of these two materials produces a hydrogel composite with remarkable strength, toughness, and lubricity.

The subject invention also pertains to a process for the surface modification of polymer surfaces to improve wettability. This process can involve the use of suitable solvents to diffuse a hydrogel precursor both into the polymer material and onto the surface. For example, a polymer with ester linkages such as polyvinyl acetate (PVAC) may be diffused into the surface of, for example PVC. Subsequent base-catalyzed or acid-catalyzed hydrolysis can convert the PVAC to PVA, improving surface wettability and enhancing bonding characteristics of the modified surface. The subject process can be employed simply to increase the wettability of the PVC surface or as an additional manufacturing step to improve the bond strength between the hydrophilic and hydrophobic components of the subject airway and medical devices.

The subject invention also pertains to a process for bonding fully hydrated PVA or other hydrogels to polypropylene, polyethylene, or other hydrophobic polymers. The subject process can involve the use of suitable solvents to allow commingling of the hydrophilic and hydrophobic surfaces while the two materials are in contact. Advantageously, co-diffusion of the two materials in contact with the solvent can allow strong bonds to develop even in the hydrated state. The subject process can be employed to securely bond hydrated PVA hydrogel

materials, for example, to PVC tubes, manifolds, or other devices using an appropriate solvent, for example dimethyl sulfoxide (DMSO).

In a specific embodiment the subject invention pertains to endotracheal tubes constructed entirely or in part of hydrogel material. Many disadvantages of the prior art endotracheal devices are overcome by the present invention, and in particular by providing an endotracheal device with a tube constructed entirely, or in part, of hydrogel material.

One object of the invention is to provide an endotracheal device that has reduced abrasiveness.

Another object of the invention is to provide an endotracheal device that can provide a nearly airtight seal with reduced pressure applied to the tracheal wall.

Another object of the invention is to provide an endotracheal device with a smooth, lubricious surface. In one aspect of the invention, the preferred surface may be provided by the hydrogel material.

Another object of the invention is to provide an endotracheal device, for example a tracheostomy tube or an endotracheal tube, that is suitable for long term, chronic applications.

In one aspect of the invention, an endotracheal device is provided with a tube constructed essentially entirely of hydrogel material. The outer diameter of the hydrogel tube swells in the presence of moisture, for example body fluid, to assist in creating a seal within the tracheal lumen, while the inner passage through the tube remains open to permit the flow of air to the patient's lungs.

In an alternative embodiment of the invention, an endotracheal device is constructed from a more rigid tube surrounded by a sleeve formed of hydrogel material. In the presence of moisture, the hydrogel sleeve can swell to create the desired seal within the tracheal lumen.

In another embodiment of the invention, a sleeve formed of hydrogel material is located over an inflatable cuff. This inflatable cuff embodiment is particularly advantageous for intubations requiring an immediate seal.

The above and other objects, advantages and features of the invention will be more readily understood from the following detailed description of preferred embodiments of the invention, which is provided in connection with the accompanying drawings.

Brief Description of the Drawings

Figure 1A is a side view of an endotracheal device constructed in accordance with the present invention.

Figure 1B is a partially broken away cross-sectional view of the endotracheal device of Figure 1A in operation, with the cross-section of the device taken along the line 2-2 of Figure 1A.

Figure 2A is a side view of another endotracheal device constructed in accordance with the present invention.

Figure 2B is a partially broken away cross-sectional view of the endotracheal device of Figure 2A in operation, with the cross-section of the device taken along the line 4-4 of Figure 2A.

Figure 3A is a side view of yet another endotracheal device constructed in accordance with the present invention.

Figure 3B is a partially broken away cross-sectional view of the endotracheal device of Figure 3A in operation, with the cross-section of the device taken along the line 6-6 of Figure 3A.

Figure 4 illustrates a generic adhesive bond of a gel sleeve to a tube in accordance with the subject invention.

Figure 5 is a graph illustrating generalized swelling kinetics of a generic hydrogel material.

Figure 6 illustrates a tracheal simulator used to conduct experiments with endotracheal devices of the subject invention.

Figure 7 illustrates performance differences between conventional cuffed endotracheal tubes and hydrogel tubes at identical ventilation parameters.

Figure 8 illustrates cell culture results from an experiment to compare a control, PVC, IPN, and PVA.

Figure 9 illustrates additional cell culture results for a control, PVC, IPN, and PVA.

Figures 10A and 10B illustrate apparatus for use in fabricating hydrogel articles in accordance with the subject invention.

Detailed Disclosure of the Invention

The subject invention pertains to novel medical devices having highly advantageous physical characteristics which reduce potential injury to patients. The excellent characteristics of the medical devices of the subject invention result, in part, from the use of hydrogel materials. A further aspect of the subject invention concerns novel manufacturing processes used to produce novel hydrogel materials and the corresponding hydrogel materials. In addition, the subject invention relates to a method for delivery of chemical substances to a patient.

The subject invention pertains to any tube, drain, catheter, or device, designed to be in contact with human tissue where the smooth, lubricious, non-abrasive, and non-adhesive surface provided by hydrogel materials would be advantageous. Furthermore, methods for delivery of drugs, chemical agents, or other substances can be accomplished utilizing the devices of the subject invention.

The subject invention pertains to medical devices which utilize hydrogels to enhance performance and reduce injuries. Hydrogels are a class of polymeric materials which swell in the presence of moisture and may contain up to approximately 95% water by weight. The high water content results in a very smooth, lubricious, and non-abrasive surface. This is an especially valuable attribute for medical devices that feature sliding contact or relative motion with living tissues.

In one embodiment the subject invention pertains to novel endotracheal tubes with tissue-contact areas constructed primarily from hydrogel materials. In a specific embodiment, an endotracheal tube device (ETD) is formed essentially entirely of hydrogel material. Alternatively, the ETD can have additional structural support in addition to the hydrogel material. In another specific embodiment, an ETD comprises a more rigid tube, for example made of PVC or other appropriate material, which is covered with a sleeve formed essentially entirely of hydrogel material, such that the reinforcement enhances the structural integrity of the ETD. Advantageously, volume swelling of the hydrogel material allows a seal to be made between the ETD and a patient's trachea. Further embodiments of ETD's employ inflatable cuffs and/or gel-cuffs to allow quick sealing of the ETD with a patient's trachea. Additional embodiments of the subject invention relate to hydrogel sleeves which can be used in conjunction with devices which contact human tissue during use, such that the use of hydrogel materials reduces the potential for patient injury.

Figures 1A and 1B show an endotracheal device (ETD) **100** incorporating a full tube design, constructed in accordance with a preferred embodiment of the present invention. Figure 1A shows a side view of ETD **100** and Figure 1B shows a cross-sectional view of ETD **100**. The device **100** includes a pliable hollow tube **102** and a manifold **150**. The tube **102** may be formed essentially entirely of a hydrogel material. The hollow tube **102** has an proximal end **104** and a distal end **106**. The proximal end **104** can be connected directly to the manifold **150**. The proximal end **104** has a central opening **108**. The distal end **106** of the tube **102** preferably has a rounded leading edge **110** for guiding the tube **102** through a patient's trachea **50**. In operation, the distal end **106** can be located within the patient's trachea **50** midway between the vocal cords and the carina.

Further, the tube **102** has a distal opening **112**, an interior passageway **114**, and an exterior surface **116**. In a preferred embodiment, the interior passageway **114** and exterior surface **116** are cylindrical. The passageway **114** connects the opposite end openings **108, 112**. Accordingly, the passageway **114** can allow gases, for example oxygen, air, or anesthetic agents, to flow freely through the device **100**.

The manifold **150**, for example a standard 15-22 mm connector, can comprise an inner passageway **152** to permit the passage of gas to and from a patient's lungs. The manifold **150** can be molded of a rigid plastic material. The manifold **150** may be releasably connected to a mechanical ventilation device.

In operation, when the tube **102** is first inserted into the trachea **50**, an annular gap **54** exists between the exterior surface **116** of the tube **102** and the inner surface **52** of the trachea **50**. Moisture within the trachea **50** can then cause the hydrogel material to swell. As a result of the swelling, the outer diameter **118** of the tube **102** increases until the exterior surface **116** comes into contact with and seals against the inner surface **52** of the trachea **50**. Thus, the moisture-induced swelling of the tube **102** closes the annular gap **54** around the endotracheal device **100**.

Since the tube **102** expands essentially along its entire length, a large area of the exterior surface **116** comes into contact with the tracheal wall **52**, and is available to form the required seal. Increasing the area of sealing contact between the device **100** and the tracheal wall **52**, by providing a relatively long portion of hydrogel material, reduces the pressure that must be applied against the tracheal wall **52** to form the desired seal.

Another advantage of the invention is that the exterior surface **116** of the tube **102** becomes smooth, lubricious and non-abrasive when wetted by body fluids within the trachea **50**. This important aspect of the device **100** reduces the possibility of injury due to tissue adhesion, abrasion, and friction.

The ETD of the subject invention can be useful in either short-term or long-term applications. The device can be inserted in a partially hydrated state, allowing the hydrogel tube to draw moisture from the humidified air and adjacent tissues to complete hydration. The uptake of moisture results in volume swelling which assists in creating a seal with the tracheal lining. In this case the seal is applied through a fluid layer at the lining surface. This seal can be distributed along the entire length of the hydrogel tube so applied local forces are minimized. Advantageously, the presence of the hydrogel material also reduces tissue trauma by decreasing friction, abrasion, applied forces, and cellular and microbial adhesion.

Another specific embodiment of the subject invention incorporates a full tube design wherein the manifold is fixed to a more rigid tube, for example made of PVC. A sleeve made

essentially entirely of a hydrogel material is then mounted around the more rigid tube. Preferably, the more rigid tube does not contact the patient's tissue, for example the trachea. This device may be employed to take advantage of volume swelling as previously described. Alternatively, the device may be inserted and employed in the fully hydrated state to take advantage of increased flexibility and enhanced surface properties. In either case, the performance of the device is improved by the smooth, lubricious, non-abrasive , and non-adhesive nature of the hydrogel surface.

Referring to Figures 2A, 2B, 3A, and 3B, the subject invention also pertains to an endotracheal tube constructed in part from hydrogel materials. A specific embodiment of the subject invention involves the placement and fixation of a hydrogel sleeve over, for example, a conventional endotracheal tube.

Figures 2A and 2B depict an ETD **300** constructed in accordance with the subject invention. Figure 2A shows a side view of the ETD **300** and Figure 2B shows a cross-sectional view of the ETD **300**. The device **300** includes a pliable hollow tube **302** with a sleeve **304** formed of hydrogel material, and a manifold **150**. The tube **302** can be, for example, made of silicone. The tube **302** may alternatively be formed of another suitable material, for example polyvinyl chloride.

The hollow tube **302** has a proximal end **306** and a distal end **308**. The proximal end **306** is connected to the manifold **150**. The proximal end **306** has a central opening **310**. The distal end **308** preferably has an angled leading edge **312** for guiding the tube **302** through a patient's trachea **50**. In operation, the distal end **308** of the tube **302** may be located, for example, within the patient's trachea **50** midway between the vocal cords and the carina.

The tube **302** has a distal opening **314**, an interior passageway **316**, and an exterior surface **318**. In a specific embodiment, the passageway **316** and exterior surface **318** can be cylindrical. The interior passageway **316** connects the opposite end openings **310**, **314** such that gas can flow freely through the device **300**.

The hydrogel material of the sleeve **304** may be the same material discussed above, which swells in the presence of moisture. The hydrogel sleeve **304** can have an proximal end **320** and a distal end **322**, and an inner surface **324** and an outer, substantially cylindrical surface **326**. The inner surface **324** of the hydrogel sleeve **304** need not be in contact with or bonded to the exterior surface **318** along the entire length of the hollow tube **302**.

The hydrogel sleeve **304** can be located near the distal end **308** of the tube **302**. The sleeve **304** can be shorter than the tube **302**, as shown in the drawings. Alternatively, the sleeve **304** can extend the entire length of the tube **302** to the proximal end **306**. The sleeve **304** is easily fabricated to lengths longer than the inflatable cuffs of prior art endotracheal devices.

This is advantageous in terms of reducing the pressure that has to be applied to the tracheal wall **52** to obtain the desired seal.

The hydrogel sleeve **304** can be attached by chemical and/or mechanical bonds **328, 330** at one or both of its ends **320, 322**. If the sleeve **304** is attached at both ends **320, 322**, the sleeve **304** may be inflated by air, water, saline, or other suitable substance, swelled with water, or both to assist in the formation of the seal against the tracheal wall **52**.

When the endotracheal device **300** is positioned in the trachea **50**, moisture within the trachea **50** can cause the hydrogel material to swell. As a result of the swelling, the outer diameter of the sleeve **304** increases such that the exterior surface **326** of the sleeve **304** comes into contact with, and seals against, the inner surface **52** of the trachea **50**. Since the sleeve **304** expands essentially along its entire length, increased area is available to form the required seal. Moreover, the wetted surface **326** of the sleeve **304** is smooth, lubricious and non-abrasive, and reduces the possibility of injury due to tissue adhesion, friction, and abrasion.

Figures 3A and 3B show an ETD **500** constructed in accordance with yet another embodiment of the present invention. Figure 3A shows a side view of the ETD **500** and Figure 3B shows a cross-sectional view of the ETD **500**. This embodiment of the ETD includes a pliable hollow tube **502** with an inflatable cuff **504**, and a manifold **150**. The tube **502** can be formed of silicone, polyvinyl chloride or another suitable material. The cuff **504** can be formed of polyvinyl chloride or other suitable material. The cuff **504** is surrounded by a sleeve **506** of hydrogel material.

The sleeve **506** may be formed of the same hydrogel material discussed above in connection with the endotracheal devices **100** and **300** of Figures 1A, 1B, 2A, and 2B. The hydrogel materials considered most advantageous for the present invention are those exhibiting high strength and elongation, minimum stiffness, and high water content.

The hollow tube **502** has an proximal end **508** and a distal end **510**. The proximal end **508** is connected to the manifold **150**. The proximal end **508** has a central opening **512**. The distal end **510** preferably has an angled leading edge **514** for guiding the tube **502** into a patient's trachea **50**, for example midway between the vocal cords and the carina.

The tube **502** also has a distal opening **516**, an interior passageway **518**, and an exterior surface **520**. In a preferred embodiment, passageway **518** and exterior surface **520** are cylindrical. The passageway **518** connects the opposite end openings **512, 516** such that ventilating gas can flow freely through the device **500**.

The hydrogel sleeve **506** is located over the cuff **504**, and has an inner surface **522** and an outer surface **524**. The inner surface **522** may contact the cuff **504** whereas the outer surface **524** can expand in the presence of moisture to form a nearly airtight seal against the

tracheal wall **52**. The hydrogel sleeve **506** can extend the full length of the tube **502** from the distal end **510** to the proximal end **508** of the tube **502** or any portion of tube **502** as desired. The hydrogel sleeve **506** is attached to the tube **502** at least at an proximal end **526** by, for example, a chemical or mechanical bond **528**.

The endotracheal device **500** can be used as follows: First, the tube **502** is inserted into the trachea **50**. Then, pressurized air, or other fluid, is supplied through an inflation conduit **530** to inflate the cuff **504**, which immediately presses the hydrogel sleeve **506** radially outwardly to form a nearly airtight seal with the tracheal wall **52**. Subsequently, moisture may cause the hydrogel sleeve **506** to swell. Then, the pressure supplied through the conduit **530** may be reduced, and the swelling of the sleeve **506** may be relied upon to maintain the seal, if desired.

Thus, the device **500** is capable of providing an immediate seal in an emergency situation, and is also suitable for long term use with reduced pressure applied against the tracheal wall **52**. The device **500** also provides reduced risk of tissue adhesion injury. As in the embodiments shown in Figures 1A, 1B, 2A, and 2B, the wetted exterior surface **524** of the hydrogel sleeve **506** is smooth, lubricious, non-abrasive, and non-adhesive.

When producing the subject hybrid tubes, it is important to attach the hydrogel sleeve securely. In a specific embodiment, a hybrid tube can be formed, for example, by joining a hydrogel sleeve to a conventional endotracheal (ET) tube. In a specific embodiment, a hybrid tube can be built on top of an 8 mm PCV Mallinckrodt ET tube. These PVC ET tubes can be purchased without cuffs or the cuffs may be removed prior to usage. Careful attention should be paid to surface regularity and cleanliness of the PVC ET tubes, regardless of the method chosen. A clean surface can facilitate the formation of a strong bond between the sleeve and the underlying tube. The bond formed may be chemical or physical, and should preferably be airtight for the sleeve to function properly. In a specific embodiment, the hydrogel sleeves can be bonded to the tube through a combination of solvent bonding and interference fit.

A specific embodiment of a hybrid tube can be made in the manner described below. A thin IPN hydrogel sleeve can be fabricated and placed over the bare tube to form a cuff dimensionally compliant to ASTM F1242-89. This assembly can be maintained in a clean, controlled environment, for example at 25°C and 45% R.H., for at least 24 hours, or until the gel has dried to a glassy finish. The sleeve can then be clamped to the tube at both ends and 2 cc of chloroform applied at the clamped PVC-IPN interface. Chloroform can be reapplied after 5, 10, and 15 minute intervals. The assembly can remain in this condition for 24 hours, after which the clamps can be removed. The bond points can be rendered waterproof by, for example, the application of silicone rubber or polyurethane spray. Shrink tubing can be placed over the

bond points as an added measure of protection against leaks. After allowing the sealants to cure the unit can be leak tested, washed, gamma sterilized, and stored in sterile distilled water.

The ETD devices **300, 500** can be useful in either short-term or long-term applications which require an immediate seal with the trachea. The gel-cuff (Type-2 hybrid), see Figures 2A and 2B, or underlying cuff (Type-1 hybrid), see Figures 3A and 3B, can be partially inflated after insertion to create an immediate seal with the lining. If the device is inserted in a partially hydrated state the hydrogel can draw moisture from the environment to enhance the seal through volume swelling. Accordingly, when the hydrogel is sufficiently swelled the cuff can be partially or totally deflated and the hydrogel can be relied upon to maintain a seal with the lining. This seal can be distributed along the entire length of the hydrogel tube so applied local forces are minimized. Alternatively, the device may be inserted in the fully hydrated state to take advantage of enhanced flexibility and surface properties. In this case the underlying cuff (Type-1 hybrid) or gel-cuff (Type-2 hybrid) can be relied upon to create and maintain the seal. The presence of the hydrogel material reduces tissue trauma by decreasing friction, abrasion, applied forces, and cellular and bacterial adhesion.

Referring to Figures 3A and 3B, a specific embodiment of the subject invention incorporating a hybrid tube design, the Type-1 hybrid tube, is shown. This embodiment can utilize a one-piece hydrogel sleeve which may or may not extend the entire length of the device. In a specific embodiment, the hydrogel sleeve is bonded to the endotracheal tube at only one location near the proximal end of an endotracheal tube, for example a conventional cuffed endotracheal tube. The underlying cuff can inflate and can be primarily responsible for achieving an initial seal between the subject device and the trachea. In addition, volume swelling of the hydrogel sleeve may augment this seal. In another embodiment the sleeve may be bonded to the endotracheal tube at both ends. This embodiment can be incorporated with an underlying cuff.

Referring to Figures 2A and 2B, another specific embodiment of the subject invention incorporating a hybrid tube design, the Type-2 hybrid tube, is shown. This embodiment may or may not extend the entire length of the device. In this embodiment, the hydrogel cuff is securely bonded to the endotracheal tube in two locations at both the machine end and the patient (distal) end of, for example, a conventional, uncuffed endotracheal tube. The hydrogel sleeve of this embodiment can comprise a true, one-piece hydrogel cuff and may be inflated with air, water, saline, or any other appropriate fluid. This embodiment can typically be employed in a fully hydrated state so the 'gel-cuff' is responsible for achieving and maintaining the seal. However, in some cases volume swelling of the hydrogel cuff may be employed to augment this seal.

The subject invention also pertains to a hydrogel sleeve or tube employed in the construction of the aforementioned hybrid and full tube designs. This tube or sleeve can be molded separately, in conjunction with, or in place over the underlying device. Advantageously, this tube or sleeve may be molded in a wide variety of cross-sectional shapes with or without a smooth surface. In a specific embodiment an imprint or shape may be molded into or imparted to the surface. This shape might serve to, for example, enhance sealing, assist mucociliary transport, or ease device extraction.

The subject invention also relates to any tube, drain, catheter, or device, with or without a cuff or balloon, designed to convey fluids into or out from the body. The subject invention can be applied to any application involving the passage of substances into or out from the body, including but not limited to airway maintenance, drug or substance delivery, dialysis, and cavity drainage. The subject invention can be applied to any device which is inserted into the body and would benefit from the smooth, lubricious, non-abrasive, and non-adhesive surface provided by hydrogel materials. Specific applications include, but are not limited to, endotracheal tubes, tracheotomy tubes, nasopharyngeal tubes, nasogastric tubes, chest tubes, wound drains, intravenous catheters, peritoneal dialysis, and Foley catheters.

In one embodiment the subject invention pertains to methods which utilize the devices of the subject invention for the delivery of chemical agents systemically or locally to specific tissues. For example, the integral hydrogel tube or sleeve can be loaded with, or bonded to, chemical agents including but not limited to antiseptics, antibiotics, anti-inflammatory agents, vectors, enzymes, and anti-thrombogenic agents. These agents can be chemically bonded to the device at any internal or external surface, or loaded into the bulk of the hydrogel. These agents may also be delivered from microspheres embedded in the device, embedded in the hydrogel sleeve, or contained within an internal cavity, This method allows complex delivery rates based on time, temperature, pH, or other factors. Referring to Figures 2A and 2B, the Type-2 hybrid design has the additional ability to deliver agents from within an internal cavity. Bulk loaded agent delivery rates may be controlled by manipulating the chemical composition of the delivered agent and hydrogel, as well as hydrogel morphology and porosity. Agent delivery rates for the Type-2 hybrid, see Figures 2A and 2B, are additionally affected by hydrogel water content, hydrogel porosity, drug concentration, and cuff pressure.

The subject invention also pertains to a process for the surface modification of polymers to improve wettability. This process can involve the use of suitable solvents to diffuse a hydrogel precursor, for example a polyester, both into the article and onto the surface. Subsequent base-catalyzed or acid-catalyzed hydrolysis can hydrolyze the polyester, improving wettability and enhancing bonding characteristics. For example, a polymer such as PVAC (polyvinyl acetate)

may be diffused into the bulk and on to the surface of a PVC article. Subsequent hydrolysis will then partially convert the PVAC to PVA, improving wettability characteristics. The subject process can be employed simply to increase the wettability of the surface or as an additional manufacturing step to improve the bond strength between the hydrophilic and hydrophobic components of the subject airway and medical devices.

The subject invention also pertains to a process for bonding fully hydrated PVA or other hydrogels to polypropylene, polyethylene, or other hydrophobic polymers. The subject process can involve the use of suitable solvents to allow interpenetration of the hydrophilic and hydrophobic surfaces while the two articles are in contact. Advantageously, co-diffusion of the two materials in contact with the solvent can allow strong bonds to develop even in the hydrated state. If desired, elevated pressures and temperatures may be employed at the bonding point to enhance the quality and strength of the bond. The subject process can be employed to securely bond, for example, hydrated PVA hydrogel tubes or sleeves to PVC tubes, manifolds, or other devices using an appropriate solvent, such as for example DMSO.

Hydrogels may be generally defined as polymers that swell in the presence of aqueous media without dissolving. This definition is expanded for purposes of the subject invention to include any polymeric material which swells in moisture but does not dissolve at physiologic temperature. This definition also encompasses any co-polymer, ter-polymer, multi-polymer, polymer blend, interpenetrating polymer network (IPN), semi-interpenetrating polymer network (SIPN) or hydrogel composite that swells in the presence of moisture and does not dissolve at physiologic temperature. This material may or may not be crosslinked. If crosslinking is desired this can be accomplished, for example, by chemical means (covalent or ionic bonds), physical entanglements, or the presence of crystallization.

In the selection and/or fabrication of hydrogel materials for use with respect to the subject invention, it is important that the hydrogel materials possess the advantageous surface properties discussed in this application. Water content is a primary factor in determining the transport, mechanical, and surface properties of hydrogel materials. Transport properties are generally determined by porosity, which is related to water content. Water also affects mechanical properties by plasticizing the gel, increasing flexibility and reducing strength. As the water content increases the surface of the material becomes much more wettable, increasing lubricity and reducing the incidence of abrasion and adhesion phenomenon. Specifically, hydrogels with high water content are generally preferred due to enhanced flexibility, wettability, and swelling performance. However, it is best to attain a balance between surface properties, which improve with increased water content, and mechanical strength, which degrades with increased water content. Accordingly, preferred hydrogel materials exhibit

superior wettability, smoothness, and lubricity, great strength, and exceptional elongation while minimizing modulus.

Hydrogels' wettability reduces the tendency of the hydrogel surface to experience cellular adhesion or bacterial colonization in aqueous environments. Cells that remain intact are less likely to suffer damage than those that adhere to a device, particularly if relative motion is involved. Cells that are damaged or removed can create a breach in the epithelium down to the basement membrane that provides a path for microbial invasion. Therefore, cellular adhesion not only increases the likelihood of primary injury, but also secondary injury through infection. Since a large percentage of hospital deaths each year are related to infection, this secondary threat may actually represent a greater potential to harm than the primary injury. The Hydrogel devices of the subject invention help to decrease the severity of this threat by reducing or eliminating the initial injury.

By placing hydrogel materials into apposition with the tracheal lining many of the injuries associated with pressure, friction, abrasion, and cellular adhesion are reduced or eliminated. Volume swelling can be employed to assist with the sealing mechanism. This strategy distributes the seal over the entire length of the device and thereby limits local applied forces. Preferably, hydrogel materials with an optimized combination of surface finish, swelling properties, and mechanical performance are utilized in accordance with the subject invention.

These material properties can be satisfied by a group of hydrogel materials including, but not limited to, poly(2-hydroxyethyl methacrylate), poly(vinyl pyrrolidone), poly(vinyl alcohol), the poly(acrylamides), and the poly(ethylene oxides). These materials can be implemented in the form of co-polymers, ter-polymers, blends, IPNs or SIPNs or hydrogel composites. If desired these materials can also be copolymerized with crosslinking agents or other monomers to increase strength.

The subject invention also pertains to a class of interpenetrating polymer network (IPN) hydrogel materials found to be particularly useful in the construction of the full tube and hybrid tube designs of the subject invention. This class of IPN materials can be fabricated, for example, from PVP (polyvinyl pyrrolidone) and PHEMA (polyhydroxyethyl methacrylate) polymers in a variety of compositions to yield a wide array of properties, including remarkable lubricity, flexibility, and toughness. Interpenetrating polymer network (IPN) materials based on poly(2-hydroxyethyl methacrylate) and poly(vinyl pyrrolidone) have proven particularly useful in the fabrication of hydrogel sleeves and tubes in accordance with the subject invention. These materials exhibit remarkable swelling, mechanical, and cellular adhesion properties.

These IPN materials can be fabricated by polymerizing a solution of poly(vinyl pyrrolidone) in hydroxyethyl methacrylate (HEMA) monomer and water. Mechanical properties

can be adjusted by modifying the component ratios. Flexibility and porosity increase with casting water content, while strength increases and flexibility decreases as the molecular weight of the PVP increases. After homogenization by mixing, these materials can be polymerized using an appropriate thermal, ultraviolet, or visible-light initiator. If desired, gamma or ion-beam radiation can also be employed to initiate the polymerization.

Poly(2-hydroxyethyl methacrylate) can be utilized in the fabrication of the subject devices. The unique properties of this material are attributed to a repeat unit which exhibits an amphiphilic nature. This feature is provided by the two substituents on the second carbon atom, a hydrophobic methyl group and a hydrophilic acrylic group. The swelling capacity of the gel is due to the highly polar nature of the carbonyl and hydroxyl functionality on the acrylic group. The elastic nature of the gel results from a combination of chain flexibility and free volume. The main chain is constructed from carbon-carbon single bonds, which enhance flexibility by allowing relatively free rotation along the entire length of the chain. The large acrylic groups increase free volume by separating the polymer chains from one another. These features account, at least in part, for the flexible nature of the material.

Poly(HEMA) hydrogels can be fabricated by bulk, solution, emulsion, or suspension polymerization. However, these materials are typically formed by casting in solution with a divinyl crosslinking agent. This method helps to prevent residual stresses in the finished product by forming the material in the swelled state. The associated free-radical polymerization reactions are normally initiated with a thermal decomposition agent such as azobisisobutyronitrile (AIBN) or benzoyl peroxide, but ultraviolet and visible light initiators, as well as gamma or electron beam radiation can also be employed. The properties of the end product can be modified by adjusting the casting water content, the degree of crosslinking, or by copolymerization with other monomers.

The casting water content refers to the mass percentage of water in the syrup, while the syrup is defined as the combination of water, monomer, polymer, initiator, and any other additives in the casting prior to polymerization. If the casting water content is lower than the EWC of the end product (approximately 35% for poly(HEMA) gels) the resulting hydrogel is normally homogenous, relatively nonporous, and optically transparent. As the casting water surpasses the EWC (from 40% - 50%) the gel begins to phase separate during polymerization. When this occurs the resulting gel becomes microporous and translucent. This structure causes the material to exhibit superior flexibility and wettability compared to nonporous gels, but tensile strength is also slightly reduced. Poly(HEMA) hydrogels fabricated with a casting water content from 50% to 60% exhibit a macroporous structure. These gels are highly wettable but are opaque and very weak. If the casting water content exceeds 60% no gel will form.

The mechanical strength of these gels can be enhanced by increasing the degree of crosslinking. This can be accomplished by, for example, increasing the amount of crosslinking agent added to the syrup. This reduces the molecular weight between crosslinks and decreases chain mobility. Unfortunately this also reduces flexibility and the ultimate degree of swelling. Because unmodified poly(HEMA) hydrogels have a fairly low EWC to begin with (about 35%) the degree of crosslinking should be kept to a minimum. Excessive crosslinking can have a negative impact on wettability and can also embrittle the gel.

Both the mechanical properties and surface characteristics of poly(HEMA) hydrogel can be modified by the addition of complementing monomers to the syrup. These monomers can possess either hydrophobic or hydrophilic character. The addition of a hydrophobic monomer such as methyl methacrylate (MMA) or lauryl methacrylate (LMA) generally has the same effect on properties as increasing the degree of crosslinking. The resulting copolymer exhibits reduced swelling, wettability, and elastic performance while tensile strength is improved. Conversely, the addition of a hydrophilic monomer such as n-vinyl pyrrolidone (NVP) or methacrylic acid (MA) improves flexibility, wettability, and swelling performance while degrading mechanical strength. As always caution must be exercised when formulating the gel to avoid poor mechanical properties or non-wetting surface characteristics. In certain situations two or three monomers in combination may be employed to tailor properties to a specific application. However, results may be unpredictable due to variations in reactivity ratios among different monomers. The following passages describe the processes employed for polymerizing poly(HEMA) hydrogels and other HEMA-based copolymers and terpolymers in accordance with the subject invention. All of these processes can employ distilled and degassed water and monomers. After mixing, the syrups can be cast immediately or refrigerated under an inert atmosphere, for example argon, for later use.

Poly(HEMA) hydrogels can be cast in a variety of vessels from, for example, an aqueous solution containing 65% HEMA monomer and 35% distilled water. The polymerization can be carried out using a 0.5% sodium bisulfite - ammonium persulfate redox initiation system. The initiator can be measured out and dissolved in the water component by swirling for several seconds. The monomer can then be added to this solution while mixing on the stirplate, preferably on low setting. After stirring, for example for about ten minutes, the solution can be poured into the molds. Unlike most other syrups, these solutions should not be stored, because the initiator can spontaneously decompose even at reduced temperature. Also, since the decomposition of the initiator is exothermic, a significant amount of heat is evolved. To prevent autoacceleration and subsequent bubble formation the castings can be chilled during polymerization. This is easily accomplished by, for example, employing an ice-water bath.

HEMA-MMA copolymer gels can be cast in a variety of vessels. In a specific embodiment, HEMA-MMA copolymer gels can be cast from a solution containing 85% HEMA monomer and 15% MMA monomer. Polymerization can be carried out using 0.5% AIBN initiator. The initiator can be measured out and dissolved in the MMA. This dissolution can be accomplished, for example, by mixing on a stirplate, preferably on low setting for about ten minutes. After the initiator dissolves, the HEMA monomer can be added and the solution can again be stirred, for example, for about an additional 20 minutes. The resulting syrup can be cast immediately, for example, by immersion in a water bath at about 45°C, or refrigerated under an inert atmosphere for later use.

HEMA-NVP copolymer gels can be cast in a variety of vessels. In a specific embodiment, HEMA-NVP copolymer gels can be cast from a solution containing 40% HEMA monomer, 30% NVP monomer, and 30% distilled water. The components can be measured out and poured together into a flask. The resulting mixture can be mixed on a stirplate, preferably on low setting for about thirty minutes. After degassing, the resulting syrup may be stored in the refrigerator under an inert atmosphere or polymerized, for example, by subjecting the casting to a 35 krad dose of gamma radiation at 25°C over an 8 hour period.

HEMA-NVP-MMA terpolymer gels can be cast in various vessels, for example, from a solution containing 40% HEMA monomer, 20% NVP monomer, 10% MMA monomer, and 30% distilled water. The components can be measured out, poured together into a flask, and mixed, for example, on a stirplate (low setting) for about thirty minutes. The syrup can be degassed and stored in the refrigerator under an inert atmosphere or polymerized, for example, by subjecting the casting to a 25 krad dose of gamma radiation at 25°C over a 6 hour period.

HEMA-NVP-LMA terpolymer gels can be cast in various vessels, for example, from a solution containing 40% HEMA monomer, 20% NVP monomer, 10% LMA monomer, and 30% distilled water. The components can be measured out, poured together into a flask, and mixed on a stirplate (low setting) for thirty minutes. The syrup can then be degassed and refrigerated under an inert atmosphere, or polymerized, for example, by subjecting the casting to a 25 krad dose of gamma radiation at 25°C over a 6 hour period.

Interpenetrating polymer networks (IPN) are normally defined as a combination of two or more intermingled polymers that are polymerized and cosslinked independently. Semi-interpenetrating polymer networks (SIPN) are similarly defined as IPN materials in which only one of the polymers is crosslinked. Both definitions are somewhat lacking because in practice crosslinking is not always necessary or even beneficial. In fact, in some cases crosslinking may actually prevent the development of desired properties. The formation of domains and entanglements is often sufficient to render the material insoluble at normal operating

temperatures so crosslinking agents can often be omitted from the syrup. For the purpose of this application IPN materials are defined to be materials composed of two or more polymeric species that are independently polymerized.

IPN systems are typically employed to enhance the properties of the 'base' hydrogel. The second component may improve the strength, flexibility, or wettability of the hydrogel depending on the concentration and chemical characteristics of the added material. In accordance with the subject invention, the addition of PVP polymer to the base poly(HEMA) hydrogel serves to increase water content, greatly improving the wettability and flexibility of the resulting material. As with other hydrogel the properties of these IPN materials can be adjusted by modifying the degree of crosslinking or the casting water concentration. However, the largest variation in performance are realized by adjusting the concentration and molecular weight of the PVP polymer phase. In a specific embodiment IPN gels employed with respect to the subject invention can be fabricated by solution polymerizing HEMA monomer in the presence of poly(vinyl pyrrolidone) polymer (10,000 - 1,000,000 M_w). The syrup is a combination of HEMA monomer, PVP polymer, distilled water, and initiator. Variation in the component ratios can be designed to tailor the gel to different applications.

A specific fabrication process in accordance with the subject invention is presented in Table 1 and involves combining two solutions, the polymer dissolved in water and the initiator dissolved in the monomer. After mixing and degassing, the resulting syrup can be cast immediately or stored under an inert atmosphere. Storage at room temperature for extended periods is safe and perhaps even beneficial. Slow polymerization of the syrup over time at this temperature can enhance the molecular weight of the HEMA phase, tending to improve the strength and durability of the resulting gel.

Table 1	A process tabulation for IPN hydrogel fabrication. PVP (300,000 M_w) can be used as received. HEMA monomer (99.5% purity) can be double distilled under vacuum. Water can also be distilled prior to use. AIBN initiator can be recrystallized from ethanol.
Step	Directions
1	Clean the mold surfaces thoroughly. After drying assemble the mold on a level surface.
2	Determine the quantities of reagent required. A specific formulation calls for a syrup composed of 10% PVP, 20% HEMA, and 70% water, such that each 100 grams of solution requires 10 grams of PVP, 20 grams of HEMA, and 70 grams of water.
3	FLASK A. Add the proper amount of water to a round bottom flask and set on a stirring hotplate. Heat the flask to 90°C.
4	FLASK B. Mix the monomer and initiator in a round bottom flask. Add an amount of AIBN equal to 0.1% of the monomer mass. Set this flask aside and mix on a stir plate (low setting).
5	Slowly add the PVP polymer to flask A. Solution is encouraged if the polymer is added slowly. Once dissolution is complete remove the heat and continue stirring. Allow the flask to cool to room temperature.
6	Add the contents of flask B to flask A. Stir at room temperature for 6 hours, then degas under vacuum for 15 minutes. The resulting solution may be stored at room temperature under an inert atmosphere or cast immediately.
7	Charge the mold to the fill line and loosely place the upper O-ring. The ring should preferably be left in this position during polymerization.
8	Place the charged mold into a heated water bath at 45°C. Maintain this temperature for 1 hour, then increase the temperature by 1°C every hour for 5 hours. Hold the temperature at 50°C for 18 hours, then increase to 60°C. Hold at this temperature for 6 hours.
9	Remove the mold from the bath and cool to room temperature. Break down the mold and remove the casting. The casting can be placed in a warm water bath to extract residual initiator and monomer.

The subject invention also relates to a process for producing PVA (polyvinyl alcohol) hydrogels found to be particularly useful in the construction of the aforementioned full tube and hybrid tube designs in accordance with the subject invention. This process is based on the well known freeze-thaw method for producing PVA hydrogels, but involves the novel use of an additional solvent to abbreviate the production cycle to a single step. This solvent is employed, at least in part, to remove water from the remaining amorphous portion of the hydrogel and allow removal from the mold. This solvent is typically a dehydrating agent such as a ketone or

alcohol, for example acetone, methanol, or ethanol. Advantageously, this process quickly produces modified freeze-thaw PVA hydrogels that are highly lubricious and very strong.

This class of polymer based on poly(vinyl alcohol) has also proven useful in the fabrication of hydrogel sleeves and tubes in accordance with the subject invention. These materials also exhibit remarkable swelling, mechanical, and cellular adhesion properties. These materials can be fabricated by gelling an aqueous solution of PVA polymer at reduced temperature, or in a mixed solvent. The mixed solvent is generally composed of a gelling agent dissolved in water. This gelling agent can be a liquid (for example dimethyl sulfoxide) or solid (for example sodium chloride) and can generally comprise up to 100 percent of the solvent. Many such gelling agents are commonly known and the polymer solution can gel at a temperature that is dependent on the type and quantity of gelling agent that is employed. The mechanical and swelling properties of the gel depend on the initial water content of the casting solution, the quantity and type of gelling agent employed, the degree of hydrolysis of the PVA polymer, and the ultimate degree of crystallization of the gel. Heat treatment, aging, or drying can be employed to adjust these properties after gelation.

Poly(vinyl alcohol) is normally produced by the acid-catalyzed hydrolysis of poly(vinyl acetate), which effectively converts the pendent acetate groups to hydroxyl groups. The properties of the resulting polymer are determined by tacticity, the degree of hydrolysis, and crosslinking. Most commercial grades of PVA are stereoregular (primarily atactic) with less than 2% of the repeat units forming in the 'head-to-head' (adjacent hydroxyl groups) conformation. This stereoregularity should allow crystallization to occur. However, crystallization can be hindered by the presence of residual acetate groups. Accordingly, the tendency toward crystallization depends primarily on the degree of hydrolysis.

The degree of hydrolysis can refer to the percentage of converted acetate groups on the main chain. Partially hydrolyzed grades (less than 75% conversion) do not crystallize significantly and are soluble in water at room temperature. This is because the large number of bulky acetate groups increases free volume and prevents the long-range interchain associations required for crystallization to occur. As the degree of hydrolysis increases the loss of bulky acetate groups reduces free volume and the chains are allowed to more closely approach one another. The compact but highly polar hydroxyl groups then come into close proximity and 'bind' the chains together through strong hydrogen bonding. These interchain forces increase the degree of crystallinity and greatly reduce solubility. In fact, in spite of the high concentration of hydroxyl groups completely hydrolyzed grades (greater than 99% conversion) of PVA must be heated to nearly 100°C to attain solution. These materials exhibit excellent

mechanical properties and chemical resistance but can also swell to a significant degree (up to 95% EWC).

The properties of PVA hydrogels can vary with molecular weight, but since these materials are normally obtained in polymer form the molecular weight cannot be adjusted. Accordingly, these properties are typically modified by means of chemical or physical crosslinking. Chemical gels can be formed by the addition of agents which undergo condensation with the hydroxyl groups on the main chain. A number of aldehydes (glutaraldehyde, formaldehyde, etc.), dicarboxylic acids (adipic acid, terephthalic acid, etc.), and metal ions (Fe^{3+}, B^{5+}, etc.) can form chemical bonds with PVA which result in crosslinks. Longer molecules such as diacids are generally preferred over ions because the metal ion 'bridge' is short and restrictive, embrittling the material. Molecules such as adipic acid can effectively restrict chain mobility while maintaining some measure of flexibility.

Physical gels can be formed by crystallization. The required orientation may be induced by drawing the material, by heat treatment, or by casting the polymer in solution with a gelling agent. These agents create specific interactions between the hydroxyl groups on adjacent chains, bringing them together to improve hydrogen bonding. Many such agents are known, and this process is easily employed on a laboratory scale.

The hydrogel sleeves fixed to the ETD devices can be cast in molds, for example as shown in Figure 10B. Figure 10B illustrates a large diameter bench top mold for use in casting PVA gels at room temperature, and Figure 10A illustrates a smaller mold for use in polymerizing hydrogel sleeves in a warm water bath. The PVA sleeve can be cast in the larger diameter mold, shown in Figure 10A, from a polymer solution in a mixed solvent. The resulting gel can form at room temperature over several hours and is not necessarily chemically crosslinked. The IPN sleeves can be cast in the smaller diameter mold, shown in Figure 10B, by polymerizing HEMA (2-hydroxyethyl methacrylate) monomer in the presence of PVP (polyvinyl pyrrolidone) polymer. The IPN sleeve can be polymerized in a water bath, so the mold has an open center channel for the passage of warm water. The upper O-ring can locate the center tube. Preferably the upper O-ring is not sealed allowing for expansion during polymerization and the evolution of gas. Attention should be paid to the fill lines to avoid contamination because the upper end of the mold is not sealed.

In a specific embodiment, the method, presented in table 2 can be employed for the fabrication of PVA gels. This process involves dissolving the polymer in a solution of water and the gelling agent, for example dimethyl sulfoxide (DMSO). This solution can spontaneously gel over several hours at room temperature or when chilled. The properties of the resulting gel depend on the molecular weight and concentration of the polymer in solution, as well as the

concentration of the gelling agent. Increasing the concentration of the agent tends to improve mechanical strength, but can also reduce EWC. The amount of gelling agent should preferably be minimized because it must be extracted prior to use.

Table 2	A process tabulation for standard poly(vinyl alcohol) hydrogel fabrication. PVA (99% hydrolyzed, 100,000 MW) and DMSO (dimethyl sulfoxide) can be used as received. Water can be distilled prior to use. The resulting syrup may be cast immediately or stored indefinitely at room temperature.
Step	Directions
1	Clean the appropriate glass mold with ethanol and assemble on a level surface.
2	Determine the quantities of reagent required. An 8% PVA solution is prepared in a 1:1 mixed solvent. 100 grams of solution will require 8 grams of PVA, 46 grams of DMSO, and 46 grams of water.
3	Create the mixed solvent by adding equal portions of water and DMSO to a round bottom flask. Set the flask on a stirring hotplate and equilibrate to 85°C.
4	Add the PVA polymer to the flask while stirring. Solution is encouraged if the polymer is added slowly. Continue stirring until dissolution is complete.
5	After solution is achieved loosely cap the flask, reduce the temperature to 75°C, and continue stirring for a period of 6 hours. After this time the solution may be stored at room temperature for later use or reheated to 85°C for casting.
6	Pour the solution into the mold and loosely place the upper seal. After any large air bubbles have escaped the seal may be replaced. If the mold cools too fast or many bubbles are trapped it may be necessary to heat the mold using a heat lamp. This decreases the solution viscosity and allows the bubbles to escape.
7	Set the mold aside and allow the casting to gel over a 24 hour period. After this time the casting may be removed and placed in a warm water bath to extract the DMSO.

Another class of polymer, modified freeze-thaw poly(vinyl alcohol) (FT-PVA), based on poly(vinyl alcohol) has also proven useful in the fabrication of hydrogel sleeves and tubes in accordance with the subject invention. These materials also exhibit remarkable swelling, mechanical, and cell adhesion properties. Typical freeze-thaw PVA hydrogels can be made by repeatedly freezing and thawing an aqueous solution of PVA. The gel begins to form as the water in the solution is removed in the form of ice crystals, allowing polymer crystals to form along parallel planes. The resulting gels are semi-crystalline and exhibit mechanical properties

that are enhanced with additional freeze-thaw cycles. The subject modified FT-PVA materials can be fabricated in a similar fashion by freezing an aqueous solution of PVA polymer. However, the process can be abbreviated by the addition of a dehydrating solvent washing step. Any appropriate water-removing solvent such as a ketone or alcohol may be employed. Some examples include, but are not limited to, acetone, ethanol, or methanol.

The aqueous PVA solution can be homogenized by mixing and then cast in a mold that is chilled in a reduced temperature bath for up to several hours. The frozen casting can then be removed from this mold and placed directly into the dehydrating solvent, which extracts much of the casting water. The mechanical properties of the resulting gel can be modified by adjusting the freezing temperature, freezing time, and solvent soak time. This procedure allows extremely strong gels to be fabricated in a shorter, single-cycle process.

Freeze-thaw (FT) PVA polymers are physical gels which can be formed by precipitating the polymer from solution. This is typically accomplished by placing the solution in a chilled bath. As this casting begins to cool, adjacent water molecules in the solution congregate and form complexes. Eventually the casting becomes supercooled and ice crystals are nucleated. When the ice formation reaches a certain point the nearby polymer chains are forcibly removed from solution and precipitated. The polymer chains in this precipitate are highly oriented and closely packed so hydrogen bonding is very effective. This allows the polymer to crystallize locally parallel to the plane of the ice crystals. The casting can then be thawed, rehydrated, and frozen once again. Subsequent freeze-thaw cycles can improve the mechanical strength of the gel by continually crystallizing the remaining amorphous material. The casting may be subjected to many more cycles, but the crystalline content will eventually reach a maximum as the amorphous material is consumed. The resulting gels can be extemely strong, highly inelastic and totally opaque. Freeze-thaw PVA polymers do not swell to the extent of conventional PVA gels but the surface is still highly wettable due to the presence of residual amorphous content. These materials also differ from standard PVA gels in that no impurities remain in the casting after gelation. The properties of these gels can be highly dependent on the initial solution concentration (porosity), freezing temperature (crystallite size), and the number of freeze-thaw cycles performed (crystalline content).

The subject invention utilizes a novel method for the fabrication of FT-PVA hydrogels. The subject method differs from the traditional freeze-thaw process and can be referred to a modified freeze-thaw method. A specific embodiment of the subject method is presented in Table 3. This embodiment of the subject method begins with a relatively slow cooling in an acetone-dry-ice bath. The nucleation and crystallization stage can be allowed to take place in this bath, for example, over about a four-hour time period. The casting can then be removed

from the mold. The casting can be immersed in, for example, liquid acetone, to remove water from the remaining amorphous material and solidify the gel. After washing the sample free of acetone the casting can be rehydrated for later use. Gels fabricated by this method differ from those produced by the standard freeze-thaw process because crystalline content is reduced. This can reduce strength but, advantageously, swelling capacity, wettability, and flexibility can be improved. More importantly, the gels can be produced in less time and at a lower cost.

Table 3 A process tabulation for modified freeze-thaw poly(vinyl alcohol) hydrogel fabrication. PVA (99.5% hydrolyzed, 100,000 MW) is used as received. Water is distilled prior to use.

Step	Directions
1	Clean and assemble the appropriate molds, for example glass molds.
2	Prepare a dry-ice acetone bath.
3	Create a homogeneous 8% PVA solution in distilled water at 85°C.
4	Pour the solution into the mold and loosely place the upper seal. After any large air bubbles have escaped the seal may be replaced. If the mold cools too fast or many bubbles are trapped it may be necessary to heat the mold, for example, using a heat lamp.
5	After cooling the solution and ensuring the removal of air bubbles, place the casting in the chilled acetone for about four hours.
6	Remove the casting from the mold and quench in acetone for one hour.
7	Wash the casting to remove residual acetone and rehydrate in distilled water.

The subject invention also pertains to a class of PVA composite hydrogels found to be particularly useful in construction of the devices of the present invention. This composite gel is created from a combination of the aforementioned modified freeze-thaw PVA hydrogels and amorphous or semi-crystalline PVA hydrogels fabricated by well-known, standard techniques. The layers or lamina of these composite hydrogels may be bonded together by means of pressure, temperature, solvent bond, chemical bond, adhesive bond, interference fit, or any combination of the above. These composite gels may be fabricated in a variety of conformations to yield a wide array of properties, including remarkable lubricity, toughness, and strength.

This class of polymer, a PVA composite hydrogel, has also proven useful in the fabrication of hydrogel sleeves and tubes in accordance with the subject invention. These materials also exhibit remarkable swelling, mechanical, and surface properties. These composite hydrogel materials are fabricated by the combination of standard PVA hydrogels and FT-PVA hydrogels, both of which have been previously described. These two materials may be layered in bulk, sheet, tube, or any other form to create a variety of intricate shapes, objects, and devices. The individual layers or lamina may be bonded by means of elevated temperature, pressure, solvent bond, adhesive bond, interference fit, or any combination of the above. The properties of the composite may be adjusted by modifying the size and properties of the individual layers, or by modifying bonding characteristics.

The hydrogel materials, for example sleeves and tubes, employed by the subject devices can be bonded to the underlying tube or manifold by various physical and chemical means. These bonding means include but are not limited to clamping, adhesive bonding, chemical bonding, solvent bonding, or interference fitting. In addition, some of these methods may be employed in combination. Furthermore, elevated temperature and/or pressure may or may not be desirable. The method selected can depend on, for example, the hydrogel material and the substrate.

Hydrogel materials can be bonded to practically any substrate with an appropriate adhesive. However, chemical interaction creates the strongest bonds. In general, any compound that wets both substrates can suffice if used in conjunction with a partial interference fit. Referring to Figure 4, an 'adhesive-interference' process has been developed to assist in bonding hydrogel materials to hydrophobic plastics. In accordance with the subject 'adhesive-interference' process, the gel sleeve can be attached to the underlying tube using, for example, an elastomeric adhesive. If desired, the dehydrated portion of the hydrogel may be covered with an additional material or coated with some substance to inhibit rehydration and protect the strength of the bond. A group of tube materials including but not limited to polyvinyl chloride (PVC), polyethylene (PE), polypropylene (PP), polyurethane (PU), natural rubber, nylon, and silicone rubber can be bonded to a group of hydrogel materials including but not limited to those previously discussed, using silicone rubber, acrylic cement, marine cement, cyanoacrylate cements, or other common adhesives. A medium-to-high molecular weight polymer or elastomer dissolved in an appropriate solvent can also make an effective adhesive.

The strength of the bond can be enhanced if both surfaces are clean and slightly roughened. The strength of the bond can also be greatly improved if the bonded section of the hydrogel material is first dried to near zero (glassy finish) water content. This eliminates the water layer at the interface and allows the adhesive to come into close proximity to both

surfaces. During dehydration, the hydrogel sleeve or tube can be placed on a flexible mandrel to allow the lumen of the hydrogel sleeve or tube to shrink to the appropriate size. Once this step is completed the adhesive can be applied to both surfaces and the components can be fitted together. A waterproof coating can be applied to the glassy portion of the gel after curing to protect the bonded region from rehydration.

In some instances it may be possible to bond the hydrogel sleeve or tube directly to the endotracheal tube or medical device with the aid of a solvent. For example, PVA hydrogels can be bonded to PVC medical devices by this method because one of the gelling agents, dimethyl sulfoxide, is also an effective solvent for the PVC. By bringing these two components together in the presence of the mutual solvent, the PVA polymer and PVC polymer are allowed to mingle at the interface. Entanglements subsequently form and eventually result in a bond whose strength is determined by the amount of solvent employed, the size of the contact area, temperature, and the contact pressure, for example, increasing any of these values will generally improve the strength of the bond.

When the water of hydration is removed, hydrogel materials become hard and glassy like many other common polymers. Increasing the equilibrium water content improves flexibility and swelling performance, but also generally decreases strength. The water content of hydrogel materials is determined primarily by chemical composition, morphology, and pore structure. In addition, there is an approximate 20 % (by mass) minimum water content to maintain acceptable wettability and flexibility. As the hydrogel dehydrates and this value is approached the material becomes glassy and difficult to manipulate. As the hydrogel swells and the water content approaches 95 % (by mass) the strength of the hydrogel decreases and may become inadequate. It is therefore important when formulating the gel to attain balance between mechanical properties and swelling properties.

When the water content is low, interchain polar interactions predominate and swelling kinetics can be poor, for example slow swelling can result. In this condition the polymer chains have a greater attraction for each other than water molecules. At high water content the polymer network density is reduced and swelling kinetics are again poor. In this state the polymer network is nearly satisfied thermodynamically so water uptake slows appreciably. Referring to Figure 5, the desired operating region occurs near the middle of the swelling curve where the uptake of water is relatively rapid. The largest volume changes experienced during swelling also occur in this area of the curve. Maximizing swelling performance is particularly important for the full tube design.

Referring to Figures 1A and 1B, a specific endotracheal device with full tube design in accordance with the subject invention utilizes a substantial, one-piece hydrogel sleeve that runs

the length of the tube. In another specific full tube embodiment of the subject invention also incorporating a substantial, one-piece hydrogel sleeve that runs the length of the tube, a centrally located tube runs the length of the device and acts as a reinforcement. This reinforcement may be constructed from PVC, polyethylene (PE), polypropylene (PP), polyurethane (PU), or any other plastic, elastomer, or metal product that has reasonable strength and flexibility. Preferably, this reinforcement will have limited tissue contact. Prior to insertion, these devices can preferably be partially hydrated with a water content ranging between about 20% and about 100% of the equilibrium water content value corresponding to maximum hydration. Once the tube is in place, referring to Figure 5, the hydrogel material can begin to swell following a path similar to the generic hydration curve. This swelling is driven by the moisture absorbed from the mucous membrane and that adsorbed from the humidified air passing through the trachea. The resulting volume change experienced during this swelling can be utilized to create a seal with the tracheal lumen. Since these tubes are essentially fluid-filled, they are incompressible and the airway pressure cannot be effectively transmitted through the tube to the tracheal wall. Accordingly, when a full tube device is used under normal conditions, the airway pressure acts only to expel the device from the trachea. Minimal forces are transmitted from the device to the tracheal wall.

An experiment was devised to test the performance of the subject full tube design endotracheal device in a simulated tracheal intubation. A realistic model must imitate not only the general shape of the trachea, but also mechanical performance and function. A specific embodiment of a tracheal model in accordance with the subject invention, as shown in Figure 6, models the three main components of the trachea, namely the super structure, the basement tissues, and the mucosal lining. With respect to this embodiment, the superstructure can be made of polypropylene rings, the basement tissues can be made of a latex-gel assembly, and the mucosal lining can be made of a hydrogel sleeve.

The latex rubber sleeve, steel keys, and machine adapter are commercially available. The polypropylene cage and patient adapter can be molded, machined from stock, or cut from tubing. The design as shown in Figure 6 employs six 1 cm wide rings equally spaced at 1 cm intervals. Variations in the number, width, and spacing of these rings are possible. The ring edges can be sanded or beveled, for example, to reduce wear on the gel lining. Once the components are assembled, the machine adapter can be sealed with silicone and the machine keys fixed with epoxy. The hydrogel sleeve is important because it forms the interface with the endotracheal tube.

Referring to Figure 6, the subject simulator was assembled using a mechanical lung and a rigid model trachea. This trachea was instrumented with an analog pressure transducer and

connected to a polygraph so that CT pressure could be measured and recorded. The simulator was then intubated and ventilated with a Bear 2000 ventilator.

Both a Type-1 PHEMA full tube endotracheal tube in accordance with the subject invention and a Mallinckrodt 6 mm ID Hi-Lo cuffed endotracheal tube were tested under identical conditions. Two experiments were performed for each tube simulating healthy lung (high compliance -0.1 L/cm H_2O) and stiff lung (low compliance -0.04 L/cm H_2O) conditions. For each experiment a tidal volume of 700 ml was delivered to the simulator at 37°C and 100% relative humidity (R.H.) with a 5% - 10% leak. The airway pressure and flow rate were monitored with a Bicore respiratory monitor while the CT pressure was measured and recorded on the polygraph. The primary objective of this experiment was to demonstrate that a seal could in fact be created using the full tube design ETD of the subject invention. A secondary objective was to compare the CT pressures for both tubes at high and low airway compliance. The results of this experiment for an ordinary cuffed 6 mm ID tube compared to a full tube in accordance with the subject invention are presented in Figure 7.

As expected a significant difference in CT pressure was demonstrated between the ordinary tube and the hydrogel tube. Referring to Figure 7, it may be seen that no pressure was transmitted to the tracheal wall for the hydrogel tube at low airway compliance, with a very small pressure increase occurring as the lung stiffened. In contrast, the cuffed endotracheal tube exerted a significant pressure on the model even at low airway compliance. In addition, the CT pressure was increased many times as the lung stiffened. This experiment suggests that the full tube design is less likely to inflict injury on the trachea than conventional cuffed tubes, especially at low airway compliance.

An experiment was devised to test the durability of the hybrid endotracheal devices and to model their effect on the trachea in a simulated intubation. A simulator was assembled similar to the simulator shown in Figure 6 using the same mechanical lung but employing a new flexible tracheal model. This model featured a soft inner PVA hydrogel lumen to imitate the mucosal membrane and serve as an interface with the tested airway devices. This lumen was designed to be a 'snap-in' component, easily replaced to avoid long delays between tests. The simulator was also upgraded with a simple linkage to provide relative motion (sliding contact) between the model and the airway device, as well as a warm water bath to keep the hydrogel model and prototypes hydrated during the experiments.

Each experiment involved intubating the simulator with either a commercially available 8mm ID Mallinckrodt Hi-Lo PVC endotracheal tube or a Type-1 IPN hybrid in accordance with the subject invention built on an identical tube. The simulator was ventilated at normal adult lung compliance with a 700 ml tidal volume while the airway was maintained at 37°C and 100%

R.H. with a 5% leak of inspiratory tidal volume. At regular intervals the experiment was temporarily interrupted so that the condition of the airway device and model could be inspected visually. Both the model and airway device were then graded on a scale from (0 - 4) for severity of damage. A zero value was given in the case of no visible damage, a value of one was assigned at the first appearance of damage, and so forth. An explanation of this grading scale is illustrated in Table 4.

Both the hybrid and conventional tubes were tested over a three day period. As shown in Table 5, during this time it is evident that the conventional tube had an immediate and pronounced effect on the simulator. Even at the relatively benign settings of normal compliance this tube quickly began to destroy the PVA sleeve in the tracheal model. Mild damage was evident at first inspection (one hour) and possibly occurred much sooner (almost immediately). This damage continued to escalate rapidly until the third day, when the first test was terminated. At this time the hydrogel sheath in the model was changed and the test restarted with the hybrid design tube.

Table 4. Description of grading scale for hybrid tube simulation.

Grade	Comments
0	No visible damage
1	Slight haziness on moist hydrogel surface
2	Mild compression or scratches
3	Deep compression, indentations, or scratches
4	Model or sleeve bursts. Complete erosion through wall.

Table 5. Results for normal-compliance simulation of an 8mm ID conventional tube and an 8mm ID Type-1 hybrid tube. Note that the lubricious nature of the hydrogel sleeve decreased friction and prevented any abrasive damage from occurring in the second test.

		Time Points (hours)						
Device	**Area**	**1**	**3**	**6**	**12**	**24**	**48**	**72**
Cuffed PVC Tube	**Tube**	-	-	-	-	-	-	-
	Model	1	1	2	2	3	3	3
Hydrogel Tube	**Tube**	0	0	0	0	0	0	0
	Model	0	0	0	0	0	0	0

The Type-1 hybrid tube in accordance with the subject invention was tested under conditions identical to the testing of the conventional tubes. The test duration remained the same and inspections were performed at the same time points. However, unlike the conventional tube, the Type-1 hybrid had no apparent effect on the tracheal model. Both the IPN sleeve on the Type-1 hybrid tube and the PVA sheath in the model were assigned zero grades for the entire duration of the test. This result illustrates the large difference between the two devices to potentially cause abrasive damage, for example, to a patient's trachea. This result also suggests that the conventional cuffed tube is much more likely to cause abrasive injury in a real tracheal environment than the hydrogel-sleeved tube.

A second simulation experiment was performed to test the durability of the Type-2 hybrid design in accordance with the subject invention and to model their effect on the trachea in a simulated intubation. The simulator was assembled as before with the flexible tracheal model and the mechanical lung set at low compliance to make the test more severe. Each experiment involved intubating the simulator with either a commercially available 8mm ID Mallinckrodt Hi-Lo PVC endotracheal tube or a Type-2 IPN hybrid design built on an identical tube with the cuff removed. The simulator was then ventilated with a 700 ml tidal volume at 37°C and 100% R.H. Every attempt was made to maintain the leak at 5%, but because of the low compliance situation the airway pressure was very high and the leak would often vary between 0% and 15%. At regular intervals the experiment was temporarily interrupted so that the condition of the airway device and model could be visually inspected. Both the model and airway device were then graded at that time for severity of damage in accordance with the grading scale of Table 4.

The experimental results, as shown in Table 6, were striking. The conventional cuffed tube caused severe damage to the model immediately and completely destroyed the hydrogel sleeve before the first 24 hour period had expired. In contrast, the Type-2 hybrid device caused only minor damage to the model, and in fact this did not even develop until near the end of the experiment. The large sleeve size and the lubricious nature of the hydrogel surface obviously provide an interface which minimizes damage.

Table 6. Results for low-compliance simulation (8 mm ID conventional and 8 mm ID Type-2 hybrid). The conventional endotracheal tube destroyed the model almost immediately, while the hydrogel tube had a very limited impact on the model and was not damaged itself.

		Time Points (hours)								
Device	**Area**	**1**	**2**	**3**	**4**	**8**	**12**	**24**	**48**	**72**
Cuffed PVC Tube	**Tube**	-	-	-	-	-	-	-	-	-
	Model	2	2	3	3	3	4	-	-	-
Hydrogel Tube	**Tube**	0	0	0	0	0	0	0	0	0
	Model	0	0	0	0	0	0	0	1	1

An experiment was devised to measure the response of living cells to several materials. This test involved aseptically placing several material disks in appropriate medium with an immortal cell line (non-fibroblastic human bronchial epithelium) and measuring the subsequent growth and morbidity of the cells. The material disks were allowed to set in this medium for various lengths of time after which they were removed and the cell wells trypsinized. The wells were then stained with methyl blue and examined under an optical microscope. Four parameters were measured for each experiment - the number of living and dead cells in the test well, and the number of living and dead cells in the control well. The results for the control and test wells were then compared.

The use of methyl blue enables a distinction to be made between living and dead cells by staining the dead cells blue. A biologically inert material is expected to have no effect on the culture, while cells will adhere to or be destroyed by a biologically active material. If a test material is inert the living and dead cell counts should be statistically similar to the control values. A large number of dead cells is a sign of material toxicity, while a large difference in (living) cell count between the test and control wells is a clear indication of adhesion phenomenon. Since the dead cells are easily identified by the blue stain it may be assumed in this case that the 'missing' cells are adherent to the material disk and removed prior to trypsinization.

The results for cell cultures on PVA hydrogel (85% water content), PVP-HEMA IPN hydrogel (60% water content), and Mallinckrodt PVC (ETT cuff material) appear in Table 7. Because the cells counted are those not adhering to the test disk it is obvious from these results

that the hydrogel materials suffer less from adhesion phenomenon than the PVC cuff material. In fact, the number of living cells in the PVC well diminish to zero almost immediately. Since the cells in this well are not being destroyed, as shown in Table 8, they must be adhering to the removed disk. In contrast, referring to Figure 8, the total number of counted living cells for both hydrogel materials more closely follow the control values. There is a statistically (t-test, α=0.05) significant performance difference at five days between the conventional PVC tube and the subject tubes utilizing hydrogel materials. The PVA hydrogel in particular appears to almost mirror the performance of the control. A seven-day total of the cell count is illustrated in Figure 9. Referring to the living cell counts shown in Figure 9, the differences between counted cells in the sample well and control well is attributed primarily to adhesion.

Table 7. Living cell count. Values are tabulated as mean ± standard deviation (n=3).

Living Cells				
Days	**Control**	**PVC**	**IPN**	**PVA**
0	10.3 ± 4.8	10.3 ± 4.8	10.3 ± 4.8	10.3 ± 4.8
1	23.0 ± 6.4	9.0 ± 6.2	5.7 ± 0.3	13.7 ± 2.1
5	158.0 ± 62.4	0.0 ± 0.0	107.7 ± 31.4	204.0 ± 44.7
7	312.0 ± 140.7	0.0 ± 0.0	71.0 ± 39.9	296.3 ± 74.3

Table 8. Dead cell count. Values are a sum of the three samples.

Dead Cells				
Days	**Control**	**PVC**	**IPN**	**PVA**
0	0	0	0	0
1	0	2	0	1
5	13	0	13	8
7	47	0	13	16

The hydrogel materials considered most advantageous for the present invention are those exhibiting high strength and elongation and minimum stiffness.

Particularly advantageous results are obtained when the hydrogel material of the endotracheal devices **100, 300, 500** is a casting of an interpenetrating polymer network (not a copolymerization) of hydroxyethyl methacrylate (HEMA) and polyvinyl pyrrolidone (PVP). In an interpenetrating polymer network (IPN), one monomer is polymerized in the presence of a polymer so that the two resulting networks are intermingled. The cast HEMA-PVP IPN has improved mechanical properties because the overall molecular weight is increased relative to a comparable copolymerization reaction.

Other hydrogel materials, including, but not limited to, modified polyvinyl alcohol, modified polyhydroxyethyl methacrylate, and random copolymers of n-vinyl pyrrolidone and hydroxyethyl methacrylate may be used with satisfactory results.

Example 1 — Full tube design.

The full tube design is divided into two subclasses, the Type-1 full tube and the Type-2 full tube. Both devices feature a substantial, one-piece hydrogel sleeve that runs the length of the tube. The Type-1 FT employs no reinforcement, while the Type-2 FT uses a centrally located tube running the length of the device. This reinforcement may be constructed from PVC, polyethylene (PE), polypropylene (PP), polyurethane (PU), or any other plastic, elastomer, or metal product that has reasonable strength and flexibility. This reinforcement does not contact tissue at any point.

Example 2— Hybrid tube design.

The hybrid tube design is divided into two subclasses, the Type-1 hybrid tube and the Type-2 hybrid tube. Both devices feature a one-piece hydrogel sleeve which may or may not extend the entire length of the device. The Type-1 HT may be bonded at one location near the machine (proximal) end of a conventional cuffed endotracheal tube or at both ends. In either case the underlying cuff inflates and is primarily responsible for achieving initial seal. In addition, volume swelling of the hydrogel sleeve may augment this seal as previously described. The Type-2 HT is bonded securely in two locations at both the machine end and patient (distal) end of an endotracheal tube device (ETD). The hydrogel sleeve on this device comprises a true, one-piece hydrogel cuff (gel-cuff) and may be inflated with air, saline, or any other appropriate fluid. These devices will typically be employed in a fully hydrated state so the 'gel-cuff' is responsible for achieving and maintaining the seal. However, in some cases volume swelling of the hydrogel cuff may be employed to augment this seal.

It should be understood that the examples and embodiments described herein are for illustrative purposes only and that various modifications or changes in light thereof will be suggested to persons skilled in the art and are to be included within the spirit and purview of this application and the scope of the appended claims.

Claims

1. A medical device, comprising a first tube, wherein said first tube comprises a hydrogel material.

2. The device, according to claim 1, wherein said first tube is open for the passage of fluids through said tube.

3. The device, according to claim 1, wherein said first tube consists essentially entirely of hydrogel material.

4. The device of claim 1, wherein said device is one of the following group: tracheostomy tube, nasopharyngeal tube, nasogastric tube, chest tube, wound drain, intravenous catheter, Foley catheter, burn treatment dressing, endotracheal tube, intravascular catheter, angioplasty catheter, peritoneal dialysis catheter, laproscope, surgical laser, endoscope, stent, guidewire and shunt.

5. The device, according to claim 1, wherein said hydrogel material holds a chemical agent such that said chemical agent is transferred to human or animal tissue during contact of the hydrogel material with the human or animal tissue.

6. The device, according to claim 1, wherein said hydrogel material holds a treatment such that said treatment is transferred to human or animal tissue during contact of the hydrogel material with the human or animal tissue.

7. The device, according to claim 1, wherein said hydrogel material holds microspheres loaded with a chemical agent such that said chemical agent is transferred to human or animal tissue during contact of the hydrogel material with the human or animal tissue.

8. The device, according to claim 7, wherein said microspheres are loaded within the bulk of the hydrogel material.

9. The device, according to claim 7, wherein said microspheres are bonded to a surface of the first tube.

10. The device, according to claim 5, wherein said chemical agents are bonded onto a surface of said first tube such that said chemical agents are transferred to human or animal tissue during contact of the hydrogel material with the human or animal tissue.

11. the device, according to claim 10, wherein said agents are bonded to a surface of the first tube.

12. The device, according to claim 1, wherein said device is an endotracheal device.

13. The device, according to claim 12, further comprising structural support means for providing structural support to the first tube.

14. The device, according to claim 13, wherein said support means comprises a material selected from the following group: thermoplastic elastomer, natural rubber, poly(dimethyl siloxanes), poly(urethanes), poly(amides), poly(vinyl chloride), poly(ethylene), poly(propylene), and stainless steel.

15. The device, according to claim 12, wherein said hydrogel material swells in the presence of moisture to assist in creating a seal or near seal with tissue of a patient.

16. The device, according to claim 15, wherein said device can be positioned in the trachea of said patient partially hydrated and assist in creating a seal or near seal with the trachea upon swelling due to moisture in the patient's airway.

17. The device, according to claim 15, wherein said seal or near seal is with a trachea of the patient.

18. The device, according to claim 15, further comprising a manifold for connection of said device to an external device.

19. The device, according to claim 18, wherein said first tube is attached to said manifold by an interference fit.

20. The device, according to claim 18, wherein said first tube is attached to said manifold by chemical bond.

21. The device, according to claim 18, wherein said first tube is attached to said manifold by an adhesive bond.

22. The device, according to claim 13, wherein said support means is a second tube positioned within said first tube.

23. The device, according to claim 13, wherein said support means comprises a spring.

24. The device, according to claim 22, wherein said second tube assists in the maintenance of patency of the device such that said device will resist collapse when introduced to an airway.

25. The device, according to claim 22, wherein said tube comprising a hydrogel material is held in place relative to the second tube by friction.

26. The device, according to claim 22, wherein a first end of said first tube is bonded to said second tube such as to hold said first tube over at least a portion of an outer surface of said second tube.

27. The endotracheal device of claim 22, wherein said second tube comprises a material selected from the group consisting of thermoplastic elastomer, natural rubber, poly(dimethyl siloxanes), poly(urethanes), poly(amides), poly(vinyl chloride), poly(ethylene), poly(propylene), and stainless steel.

28. The device, according to claim 22, wherein a cavity between said second tube and said first tube can be loaded with a chemical agent such that said chemical agent passes through the hydrogel material of the sleeve and is transferred to human or animal tissue during contact of the hydrogel material with the human or animal tissue.

29. The device, according to claim 22, wherein said first tube is bonded to said second tube with an adhesive selected from the group consisting of thermoplastic elastomer, elastomeric adhesive, acrylic adhesive, cyanoacrylate adhesive, poly(urethane) based adhesive, poly(dimethyl siloxane) based adhesive, and epoxy type adhesive.

30. The device, according to claim 29, wherein a coating is applied over the adhesive and the portion of the hydrogel material in contact with the adhesive, thereby protecting the bond from weakening due to rehydration.

31. The device, according to claim 29, wherein said coating is a polymer selected from the group consisting of thermoplastic elastomer, natural rubber, poly(dimethyl siloxane), poly(urethanes), and epoxy.

32. The device, according to claim 26, wherein a second end of said first tube is bonded to said second tube.

33. The device, according to claim 22, further comprising an inflatable cuff for expanding said first tube, said cuff being located between said first tube and said second tube.

34. The device, according to claim 32, wherein said first tube can act as a cuff such that a fluid may be introduced between said first tube and said second tube.

35. The device, according to claim 1, wherein said hydrogel material comprises a material selected from the group consisting of poly(hydroxyethyl methacrylate), poly(vinyl pyrrolidone), poly(vinyl alcohol), poly(ethylene oxide), and poly(acrylamide).

36. The device, according to claim 1, wherein said hydrogel material comprises an interpenetrating polymer network.

37. The device, according to claim 1, wherein said hydrogel material comprises a semi-interpenetrating polymer network.

38. The device, according to claim 36, wherein said interpenetrating polymer network comprises poly(hydroxyethyl methacrylate) and poly(vinyl pyrrolidone).

39. The device, according to claim 38, wherein said interpenetrating polymer network is formed by polymerizing hydroxyethyl methacrylate monomer in the presence of poly(vinyl pyrrolidone) polymer.

40. The device, according to claim 1, wherein said hydrogel material comprises a modified freeze-thaw PVA.

41. The device, according to claim 40, wherein said modified freeze-thaw PVA is produced by cooling a sample of polyvinyl alcohol hydrogel below the normal freezing point of water until the polyvinyl alcohol acrystals precipitate and subsequently applying a dehydrating solvent to the sample.

42. The device, according to claim 1, wherein said hydrogel material comprises a hydrogel composite.

43. The device, according to claim 42, wherein said composite is a laminate of more than one hydrogel materials.

44. The device, according to claim 43, wherein said hydrogel composite is a laminate of at least one soft, lubricious hydrogel and at least one stronger, tougher, structural hydrogel.

45. The device, according to claim 43, wherein said hydrogel composite comprises: at least one hydrogel selected from the group consisting of poly(vinyl alcohol), poly(ethylene oxide), and poly(acrylamide); and at least one hydrogel selected from the group consisting of hydrophilic silicone rubber, hydrophilic poly(urethane), poly(acrylonitrile) based hydrogel, freeze-thaw poly(vinyl alcohol), and modified freeze-thaw poly(vinyl alcohol).

46. The device, according to claim 43, wherein said laminate comprises a first hydrogel material near an inner portion of said first tube and a second hydrogel material near an outer portion of said first tube, wherein said first hydrogel material is stiffer than said second hydrogel material and said second hydrogel material is more lubricious in the presence of moisture than said first hydrogel material.

47. The device, according to claim 15, wherein said first tube comprises a cross-sectional shape selected from the group consisting of circular, semicircular, elliptical, semi-elliptical, and triangular, wherein said cross-sectional shape facilitates creating said seal or near seal.

48. The device, according to claim 15, wherein an exterior surface of said first tube comprises a texture selected from the following group: undulated, corkscrewed, and ribbed; wherein said texture facilitates mucosal transport.

49. The device of claim 15, wherein said first tube has a shape imparted to an exterior surface of said first tube wherein said shape facilitates the creation of a seal or near seal with tissue of a patient or device insertion and extraction.

50. A method of intubating a patient, said method comprising the steps of:

providing an endotracheal device comprising a first tube, wherein said first tube comprises a hydrogel material;

locating said endotracheal device within a trachea of said patient and permitting moisture to come into contact with said hydrogel material, such that said hydrogel material swells to create a seal or near seal with said trachea; and

flowing fluid through said endotracheal device.

51. The method of claim 50, wherein said endotracheal device further comprises a manifold, further comprising the step of connecting said manifold to a ventilator.

52. The method of claim 50, wherein said hydrogel material comprises a hydrogel composite.

53. The method of claim 52, wherein said hydrogel composite comprises poly(vinyl alcohol) and modified freeze-thaw poly(vinyl alcohol).

54. The method of claim 50, wherein said hydrogel material comprises an interpenetrating polymer network.

55. The method of claim 50, wherein said hydrogel material comprises a semi-interpenetrating polymer network.

56. The method of claim 54, wherein said interpenetrating polymer network comprises hydroxyethyl methacrylate and polyvinyl pyrrolidone.

57. A process for producing a polyvinyl alcohol hydrogel, comprising the following steps:

freezing an aqueous solution of polyvinyl alcohol; and

applying a solvent to the sample.

58. The process, according to claim 57, wherein said solvent is a dehydrating agent.

59. The process, according to claim 58, wherein said solvent is selected from the group consisting of aldehyde, ketone, alcohol, formaldehyde, glutaraldehyde, acetone, methanol, ethanol, and propanol.

60. The process, according to claim 57, further comprising the step of thawing the sample.

61. A method for modifying the surface of a polymer object, comprising the following steps:

applying a first solvent to the surface of an object comprising a first polymer;

applying a solution of a hydrogel precursor of a second polymer in a second solvent to the surface of the object; and

hydrolyzing the hydrogel precursor,

wherein applying the solution of the hydrogel precursor of the second polymer in the second solvent deposits the hydrogel precursor into the bulk and onto the surface of the object.

62. The method, according to claim 61, wherein the step of hydrolyzing the hydrogel precursor comprises a process selected from the group consisting of base-catalyzed hydrolysis and acid-catalyzed hydrolysis.

63. The method, according to claim 61, wherein the first and second solvents and the first and second polymers are selected so that the associated solubility parameters thermodynamically favor deposition of the second polymer onto the first polymer and the diffusion of the second polymer into the first polymer.

64. The method, according to claim 63, wherein the first and second solvents are selected from a group consisting of acetone, propanol, methanol, ethanol, tetrahydrofuran,

cyclohexanone, dimethyl formamide, pyridine, glutaraldehyde, formaldehyde, dimethyl sulfoxide, epichlorohydrin, and chloroform.

65. The method, according to claim 61, wherein said hydrogel precursor contains ester functionality.

66. The method, according to claim 65, wherein said hydrogel precursor is selected from the group consisting of poly(vinyl acetate), poly(t-butyl acrylate), and cellulose acetate.

67. The method, according to claim 64, wherein said hydrogel precursor is selected from the group consisting of (poly(vinyl acetate) and poly(t-butyl acrylate)), poly(vinyl acetate), poly (t-butyl acrylate), and cellulose acetate.

68. A process for bonding a hydrogel object to a hydrophobic polymer object, comprising the following steps:

bringing the surfaces of a hydrogel object and a hydrophobic polymer object into contact; and

applying a solvent to the contacting surfaces,

wherein the solvent allows interpenetration of the surfaces in contact.

69. The process, according to claim 68, wherein the hydrogel object is essentially fully hydrated.

70. The process, according to claim 68, wherein the solvent causes the hydrogel object to swell.

71. The process, according to claim 68, wherein the hydrogel object is selected from a group consisting of poly(vinyl alcohol), freeze-thaw poly(vinyl alcohol), and modified freeze-thaw polyvinyl alcohol).

72. A sleeve for incorporation into a medical device, comprising an essentially tubular body of hydrogel material, wherein said sleeve is adapted to attach to a body of a medical device.

73. The sleeve, according to claim 72, wherein said medical device is selected from the group consisting of endotracheal tube, tracheotomy tube, nasopharyngeal tube, nasogastric tube, chest tube, wound drain, intrvenous catheter, intravascular catheter, peritoneal dialysis catheter, urinary catheter, embolectomy catheter, occlusion catheter, irrigation catheter, angiography catheter, angioplasty catheter, guidewire, carotid shunt, stent, and surgical laser device.

74. The sleeve, according to claim 72, wherein said medical device benefits from the biocompatible, smooth, lubricious, and non-adhesive surface provided by the hydrogel sleeve.

75. The sleeve, according to claim 72, wherein said hydrogel is selected from the group consisting of a hydrogel composite, interpenetrating polymer network, or semi-interpenetrating polymer network.

76. A method for producing a hydrogel article, comprising the steps of:

creating a mixed solvent comprising water and solvent by combining water and solvent in a container and mechanically mixing the water and solvent at a temperature between about 40°C and 105°C;

adding a polymer to the container while stirring and continuing stirring until dissolution of the polymer is essentially complete;

capping the container, reducing the temperature to between about 40°C and 90°C, and continuing stirring between about 0.5 hours and 12 hours;

reheating the solution to between about 40°C and 90°C;

pouring the solution into a mold;

sealing the mold;

allowing the solution to gel; and

exposing the gel to water in order to extract residual solvent.

77. The method, according to claim 76, wherein said solvent is selected from the group consisting of DMSO, water, methanol, ethanol, propanol, acetaldehyde, formamide, and 2-pyrrolidine.

78. The method, according to claim 77, wherein said water and said solvent are combined in approximately equal portions by weight.

79. The method, according to claim 76, wherein the mechanical mixing of the water and solvent in the step of creating a mixed solvent is conducted at a temperature of about 85°C, and wherein the solution is reheated to about 85°C.

80. The method, according to claim 76, wherein said polymer is a PVA polymer.

81. The method, according to claim 79, wherein after capping the container, the temperature is reduced to about 75°C and stirring is continued for about 6 hours.

82. The method, according to claim 76, wherein said mold comprises concentric glass containers separated by a first o-ring which essentially seals the bottom of the mold and wherein said mold is sealed with a second o-ring.

83. A method for producing a hydrogel article, comprising the steps of:

adding a polymer to a first container of water, wherein said water is being mechanically mixed at a temperature between about 40°C and 105°C;

allowing the solution in the first container to cool to ambient temperature;

mixing a monomer and an initiator in a second container, wherein an amount of initiator equal to between about 0.1 % and about 5 % of the monomer mass is mixed with the monomer;

adding the contents of the second container to the first container and continuing to mechanically mix the contents of the first container for between about 0.5 hours to about 24 hours;

degassing the contents of the first container under vacuum for between about 5 minutes and about 60 minutes;

pouring the contents of the first container into a mold;

loosely sealing the mold;

placing the mold in an environment between about 35°C and about 75°C for between about 0.5 hours and about 24 hours, such that a gel is formed;

removing the gel from the mold; and

exposing the gel to water in order to extract residual initiator and monomer.

84. The method, according to claim 83, wherein said polymer is PVP polymer, said monomer is a HEMA monomer, and said initiator is a AIBN initiator.

85. The method, according to claim 84, wherein said PVP polymer is added to the first container of water at a temperature of about 90°C.

86. The method, according to claim 84, wherein the amount of AIBN initiator mixed with the HEMA monomer is equal to about 0.1% of the HEMA monomer mass.

87. The method, according to claim 84, wherein the HEMA monomer and the PVP polymer are in a ratio between about 1:1 and 5:1.

88. The method, according to claim 84, wherein the ratio of the amount of water in the first container: the amount of PVP polymer added to said water: the amount of HEMA monomer is about 70:10:20, such that each 100 grams of solution requires about 70 grams of water, 10 grams of PVP polymer,and 20 grams of HEMA monomer.

89. The method, according to claim 84, wherein after the mold is loosely capped, the environment the mold is placed in is a heated water bath.

90. The method, according to claim 89, wherein the temperature of said heated water bath is maintained at about 45°C for about 1 hour, then increased by about 1°C every hour for about 5 hours, then held at about 50°C for about 18 hours, then increased to about 60°C for about 6 hours.

91. A device for modeling a trachea, comprising a hydrogel sleeve to simulate a mucosal lining of a trachea.

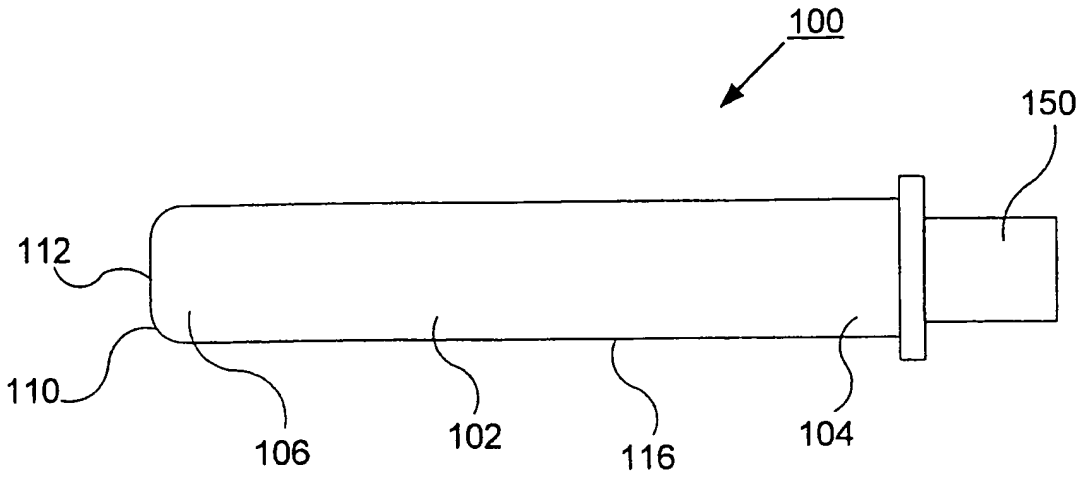
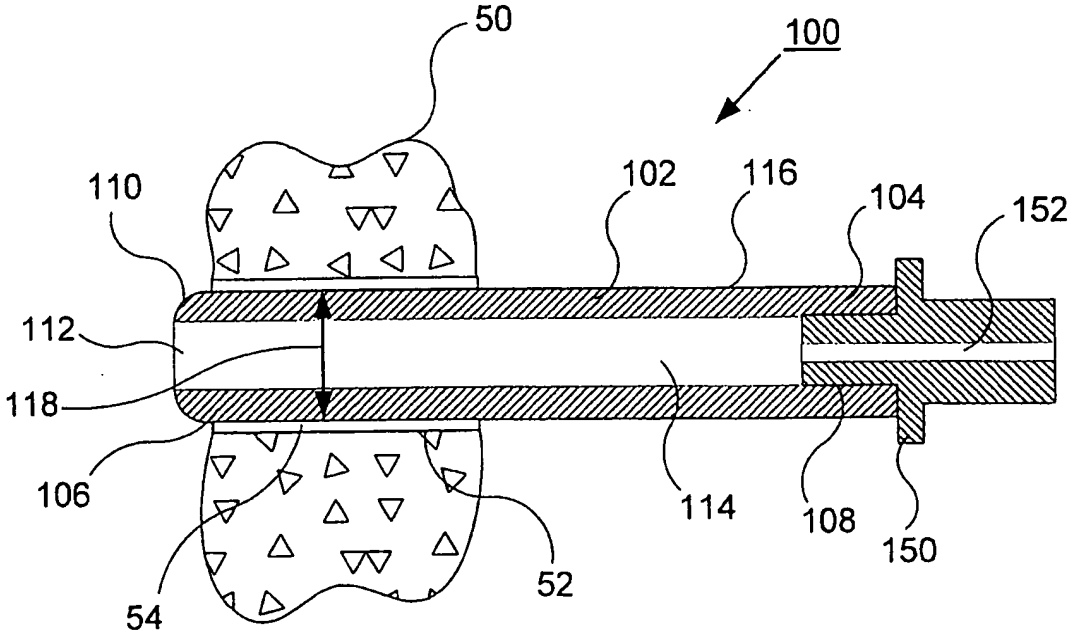

100
150
112
110
106
102
116
104

FIG. 1A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114
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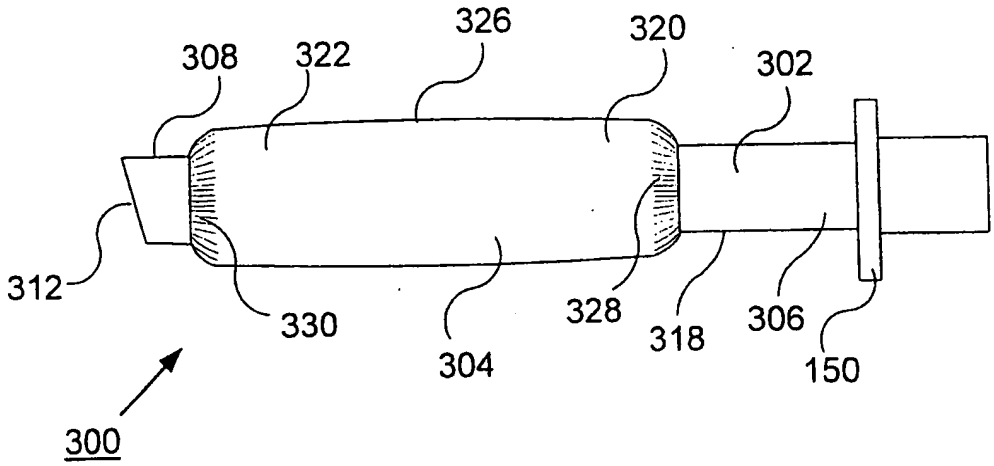
54
52

FIG. 1B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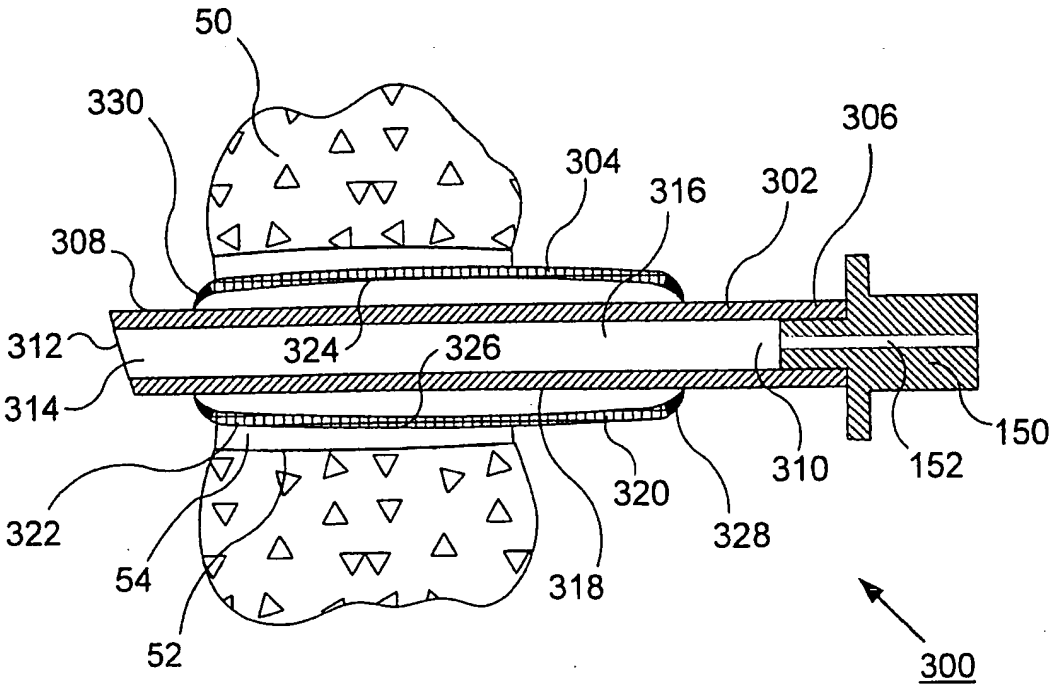
320
308
322
302
312
330
328
318
306
304
150
300

FIG. 2A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330
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322
328
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318
52
300

FIG. 2B

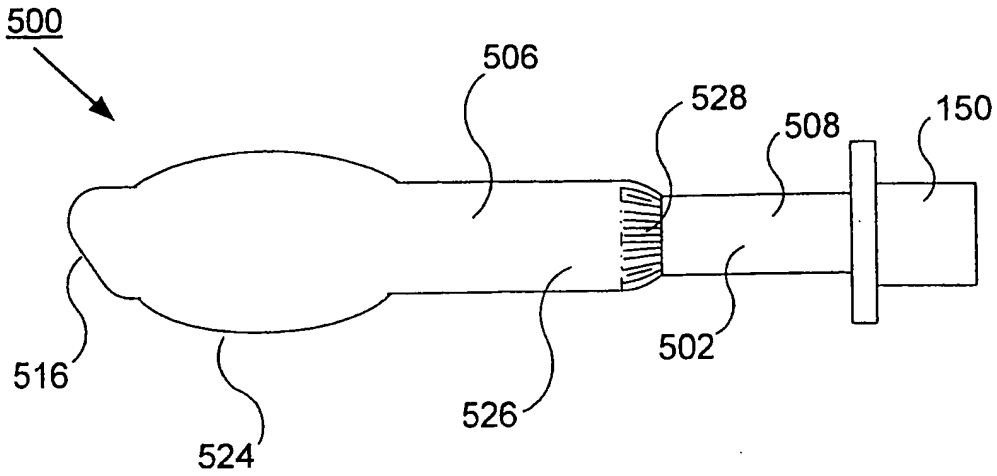
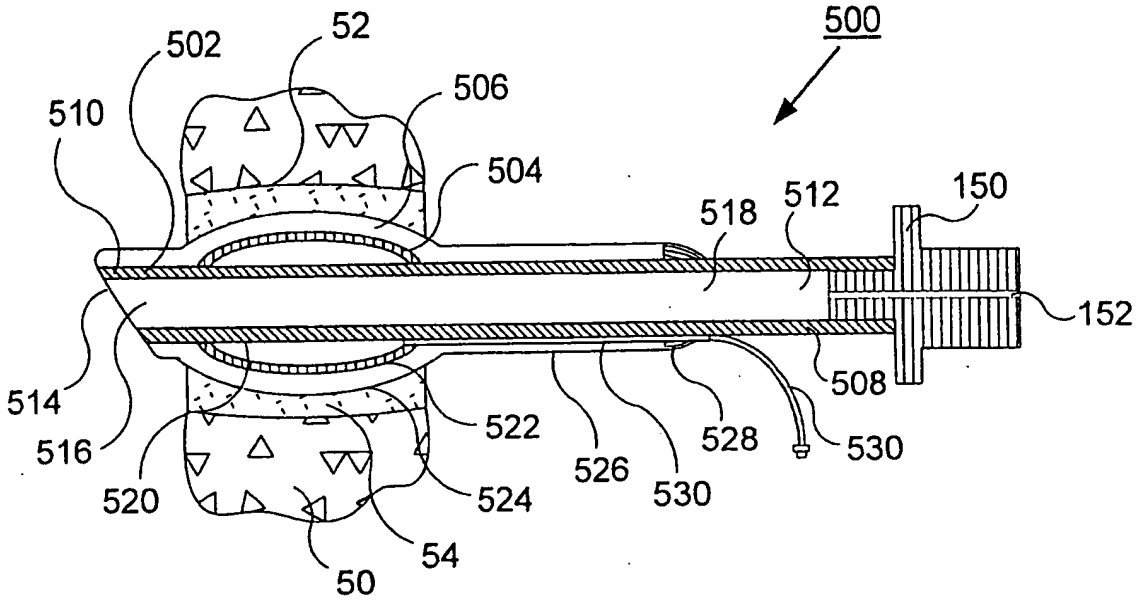

500
506
528
508
150
516
524
526
502

FIG. 3A


502
52
500
510
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150
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520
522
524
526
530
528
530
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54
50

FIG. 3B

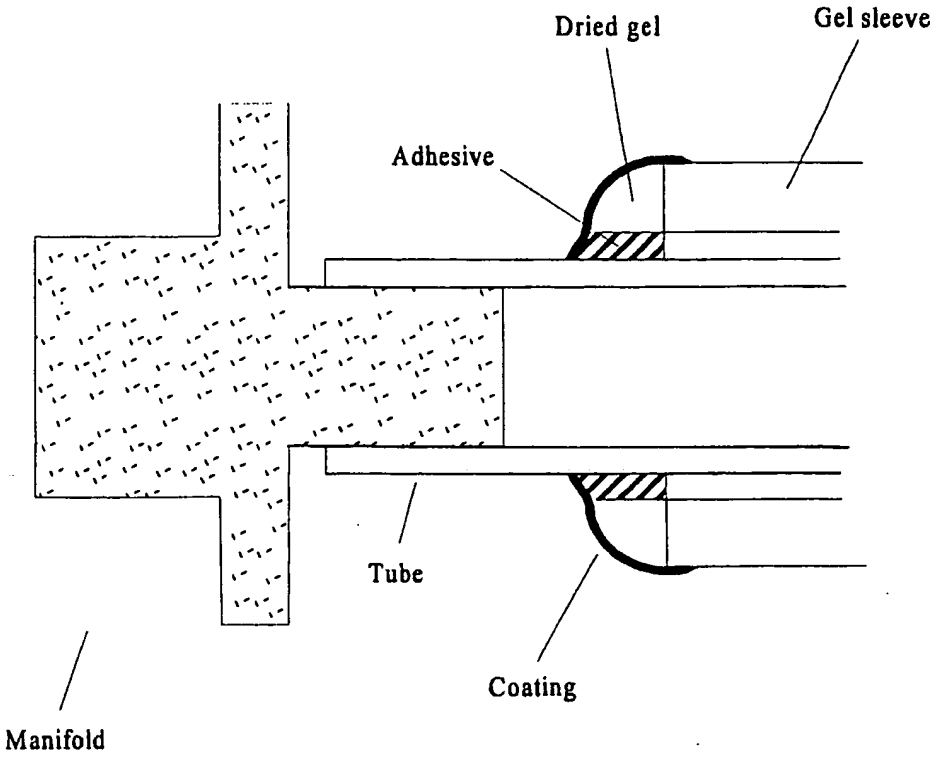


FIG. 4

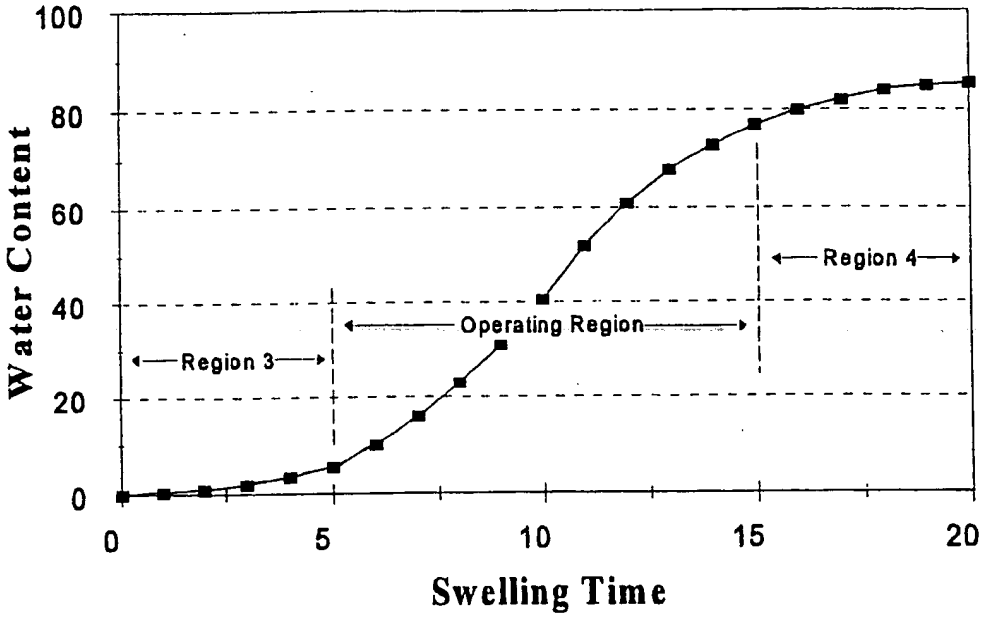
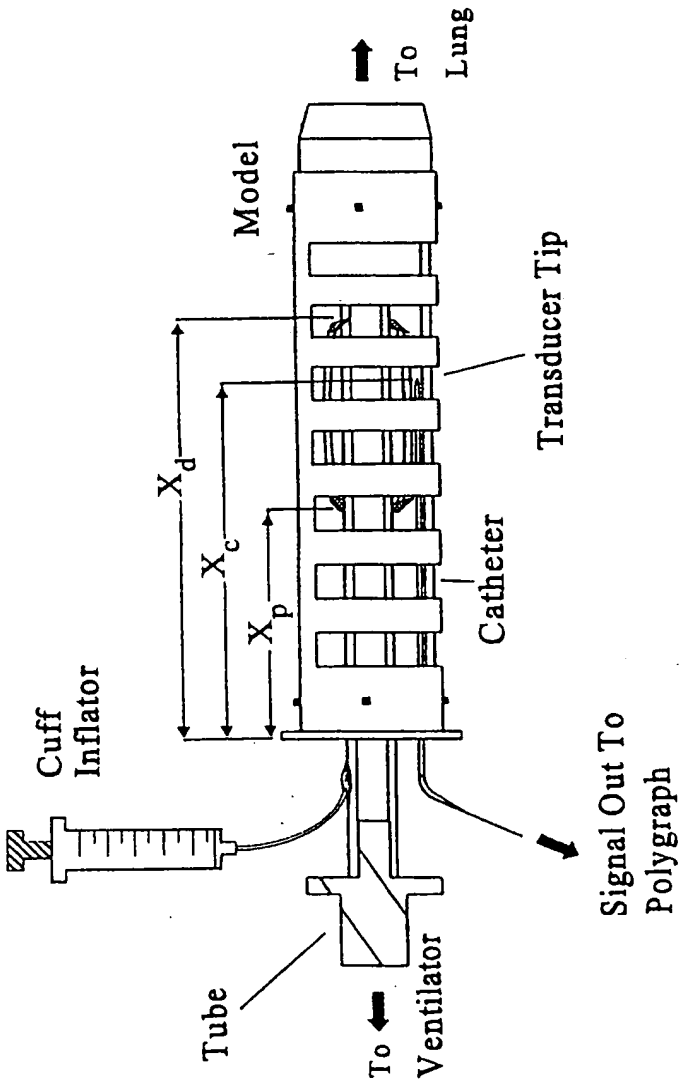

100
80
60
40
20
0
Water Content
0
5
10
15
20
Swelling Time
Region 3
Operating Region
Region 4

FIG. 5

FIG. 6A



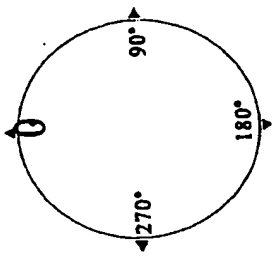


FIG. 6C

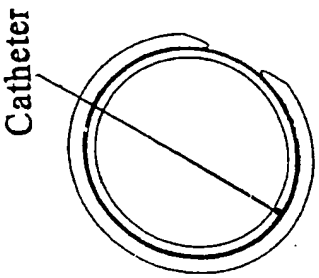


FIG. 6B

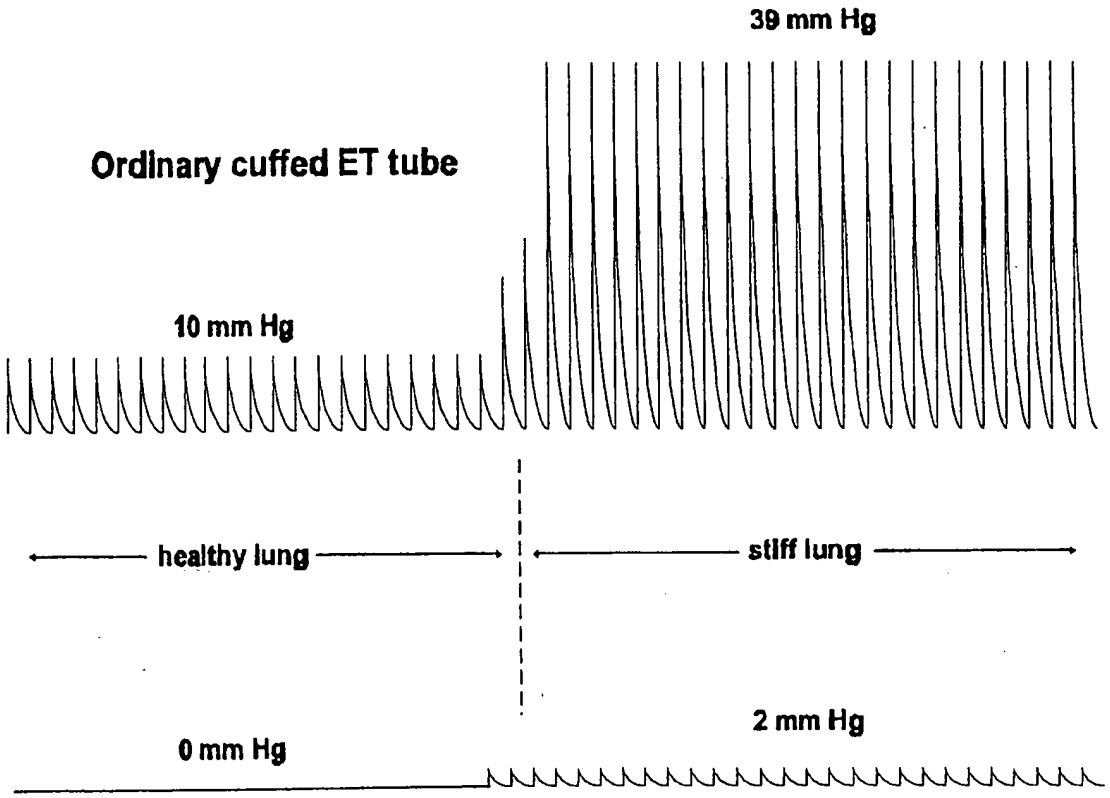

39 mm Hg
Ordinary cuffed ET tube
10 mm Hg
healthy lung
stiff lung
2 mm Hg
0 mm Hg
Full tube prototype

FIG. 7

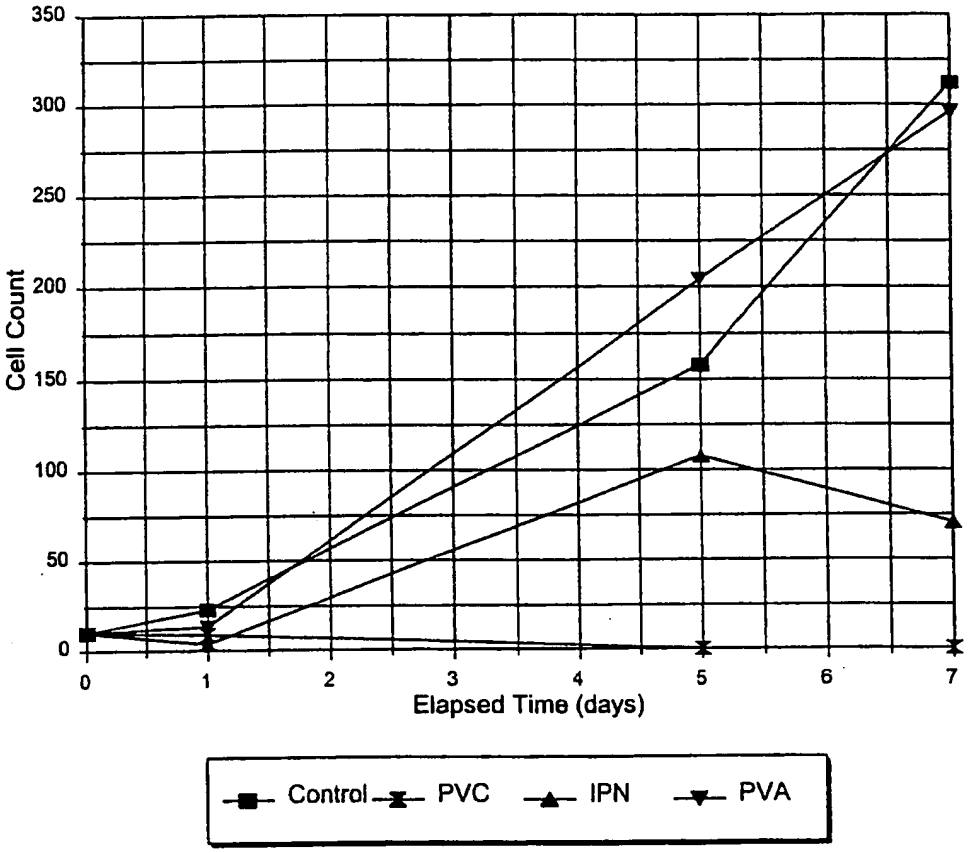
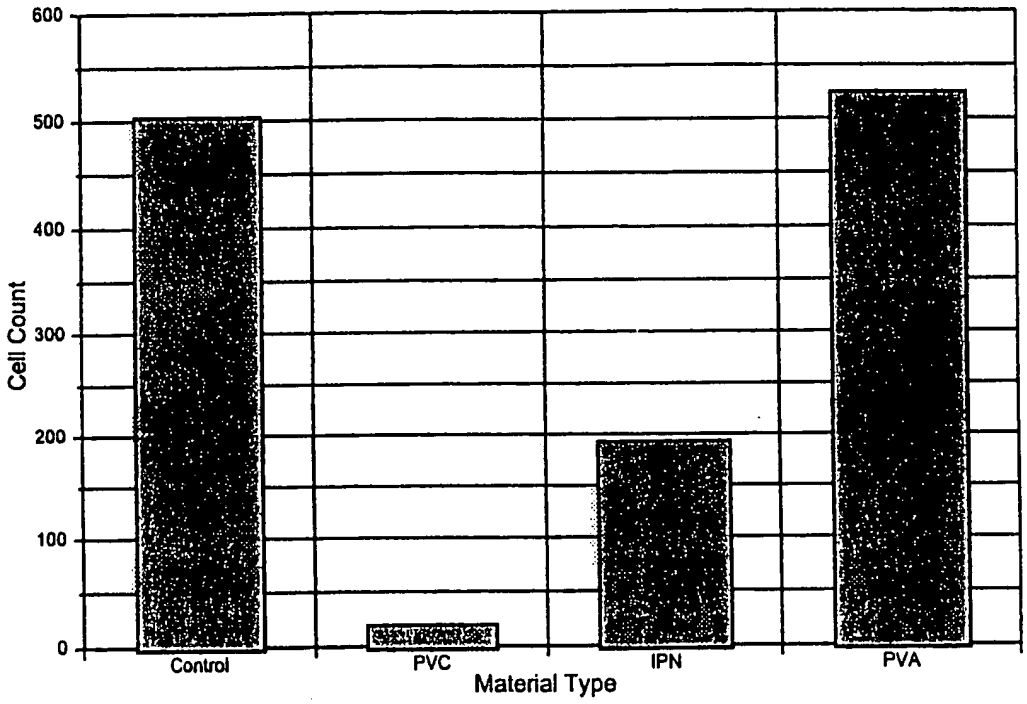

350
300
250
200
150
100
50
0
Cell Count
0
1
2
3
4
5
6
7
Elapsed Time (days)
Control
PVC
IPN
PVA

FIG. 8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500
400
300
200
100
0
Cell Count
Control
PVC
IPN
PVA
Material Type

FIG. 9

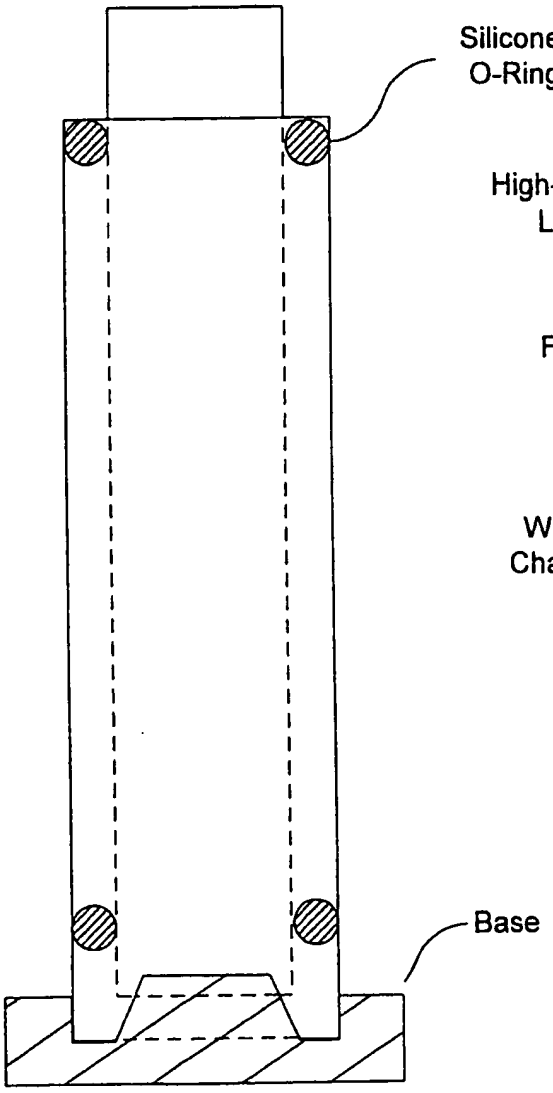

FIG. 10A

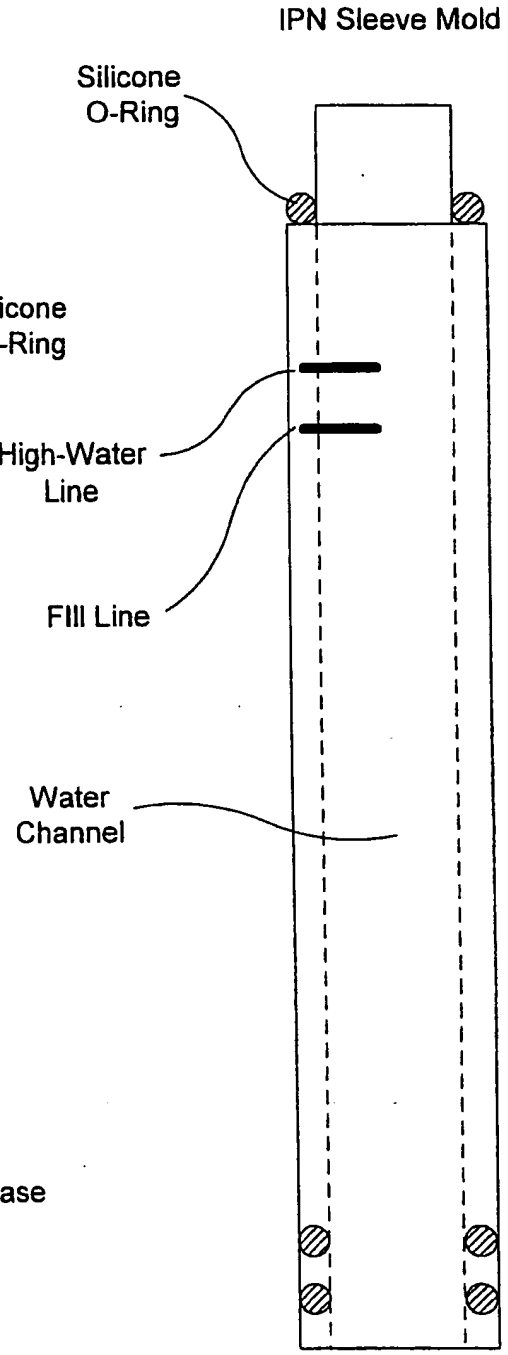

FIG. 10B